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Impact Acceleration Stress

Proceedings of a Symposium
With a Comprehensive Chronological Bibliography

National Academy of Sciences—
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Publication 977

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George Derbyshire
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Impact Acceleration Stress

A Symposium

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With a Comprehensive Chronological Bibliography

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PREFACE

The symposium on Impact Acceleration Stress was held November 27-29, 1961 at Brooks Air Force Base, Texas, under the auspices of the Man in Space Committee of the Space Science Board, National Academy of Sciences. Support for the conference was provided by a grant from the National Aeronautics and Space Administration; transportation assistance for most of the foreign participants was provided by the Departments of the Air Force and Navy. Chairman of the conference was Dr. James D. Hardy, Yale University; vice-chairman was Dr. Carl C. Clark, Martin-Marietta Corporation. Space Science Board staff support was provided by Mr. George A. Derbyshire and Thomas Gikas. Editorial review of the papers was performed by Mr. Robert Hume, NAS-NRC Publication Editor and his assistance in this regard is gratefully acknowledged.

Senior representatives of the Departments of the Army, Air Force and Navy, National Aeronautics and Space Administration, Federal Aviation Agency, U.S. educational institutions and laboratories and industrial organizations and representatives from foreign laboratories actively participated in the symposium sessions. The program included presentation of formal papers and three panel discussions. In some instances panel participants submitted formal papers and these have been included in the proceedings. Some participants listed in the program (Appendix 2) were unable to be present and did not forward a paper subsequently for inclusion in the completed publication. In other instances, participants advised that their papers would be printed in appropriate scientific journals and requested the avoidance of a duplication of effort.

A bibliography has been included with this publication to provide readers with fairly comprehensive information on literature relating to the biological effects of impact acceleration. It represents a preliminary effort to develop a more complete bibliography on this subject and readers are encouraged to forward additions and corrections to the authors.

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WELCOME

Brigadier General T. C. Bedwell, Jr.
Commander, Aerospace Medical Division
Brooks Air Force Base, Texas

It gives me a great deal of pleasure, as Commander of the new Aerospace Medical Division of the Air Force Systems Command, to join Dr. Hardy in making you welcome at this timely Symposium on Impact Acceleration Stress. The fact that your committee chose to have the meeting here is extremely gratifying to all of us in the Aerospace Medical Center and the School of Aerospace Medicine.

The problems of acceleration stress and injury have always been important to the Air Force. Our vehicles—even the simplest ones—operate at relatively high velocities. They seek to maintain a delicate balance of forces in a three-dimensional medium that permits many complex kinds of motion.

The basic art of flying is easily learned. But perhaps it is not so easily understood by nerves that react to varying conditions in flight. So a large part of our study, in aerospace medicine, is concerned with changes in acceleration, and with their effects on the human body.

Often the changes are radical, sometimes they are abrupt, and occasionally they turn out to be disastrous. Considering the peculiar nature of the equilibrium at which we aim, it is remarkable that aviation—and especially military aviation—has achieved such a high degree of safety.

Today, our medium is being extended in all directions, toward the immensity of space. The velocities attained by our vehicles have been multiplied exponentially. Unfamiliar forces are creating motions of a new order of complexity. In these circumstances, the effects produced by rapid changes in acceleration are potentially more serious than ever before in the history of human flight.

I assume that it is this background of research in all the aspects of acceleration—including impact, with its causes, its tolerances, and its pathology—which has led the space committee of the Space Science Board to bring the present symposium to Brooks Air Force Base.

The Aerospace Medical Center, as many of you know, is the oldest and the most complete facility in the Air Force for medical research, practice, and teaching. I say this without meaning to be boastful, but simply as a statement of fact.

Since 1918, we have been investigating the whole spectrum of conditions which affect the health and efficiency of men in flight, whether through the atmosphere or through space. From our studies, we develop therapeutic standards for the Air Force Medical Service. We apply them in cases which are referred to us for consultation or treatment.

The concept of space medicine, as a systematic investigation of the biological environment for man in space, was created here more than a dozen years ago. Many other similar laboratories, for research in specialized areas, have appeared in recent years. Some are sponsored by the aerospace industry, some by civilian institutions, and some by other agencies of the government.

Yet I think I am correct in saying that the Aerospace Medical Center continues to have the broadest capability in this field. I believe it is also true that the great majority of these biological problems were explored first by our Department of Space Medicine. Hence, our research staff is blessed with a wide range of practical experience.

Just four weeks ago, the Center was transferred from the Air Training Command to the Air Force Systems Command of General Bernard A. Schriever. It is now the Aerospace Medical Division of that command, which is responsible for the development of military satellites and space craft, ballistic missiles, and other advanced military aerospace systems.

The Center takes with it, into the Systems Command, the facilities it now has Brooks and at Lackland Air Force Base, a short distance away. They include—besides the School of Aerospace Medicine—the modern Air Force Hospital at Lackland.

In addition, the new Division will receive several valuable accretions in other places. Among them are the Aerospace Medical Research Laboratories at Wright-Patterson Air Force Base, Ohio; the Arctic Aeromedical Laboratory at Fort Jonathan M. Wainwright, Alaska; and the Aeromedical Research Laboratory at Holloman Air Force Base, New Mexico.

The laboratory at Holloman is of particular interest to the group meeting here today. It was there that Colonel John Paul Stapp conducted a series of vitally significant experiments in rapid deceleration, using a rocket-propelled sled on the high-speed test track. A number of you will be visiting Holloman on Thursday, after the formal program is concluded.

The object of the reorganization now in progress is of the utmost importance in our national space effort. It is to bring together, under one management, the most outstanding Air Force activities which are carrying on space research programs in the life sciences. For some time, we have been aware that there was a considerable amount of duplication in these programs and facilities within the Air Force as well as in other military and civilian agencies. The consolidation of all these laboratories, under unified direction, will put an end to the replication of programs within the Air Force.

For the Systems Command, the addition of the medical facilities will provide direct capability, within the Command, to guide and support the development of manned space systems. For the Center, the association of our research with a large operational and engineering organization will provide immediate access to experimental resources—for example, satellites and space craft—which have been available only by invitation before.

In effect, this means that bioastronautics research in the Air Force will now experience an acceleration of its own, to match the quickening of progress in the physical sciences and technology over the past decade. Hitherto, we have managed to stay

well ahead of the advances in vehicle design for manned space flight. But that was because the mission as a whole remained largely theoretical.

Since April of this year, when Yuri Gagarin orbited the earth for the first time, space flight has been a theoretical project no longer. Today, we have an urgent need for space vehicles, in which to verify and expand the data we have obtained so far almost entirely by means of simulated techniques on the ground.

For the accelerated effort in space medicine that lies ahead, the Air Force is fortunate to have a substantial base already in existence, here at Brooks. The half-dozen modern buildings you have seen this morning on your way to the auditorium where you are now meeting have been completed and equipped within the last three years. They were erected at a cost of more than ten million dollars to house the aerospace medical programs which we are conducting at this time.

You could hardly have failed to notice the five additional structures now going up. The noise of earth-moving machines, cranes, and riveters would make the discussions we are carrying on here highly impractical except for the fact that we are in a sound-proof hall.

With these additional laboratories and offices, plus the ones we are acquiring at Wright-Patterson, Fort Wainwright, and Holloman, the Aerospace Medical Division will have an early capability to manage the space-research requirements of the nation—in the life sciences—for the next five or ten years.

There will be no necessity to reserve this capability for the sole use of the Air Force. Our physical plant, our technical resources, and our professional staff will be able to assist in the research needs of the National Aeronautics and Space Administration, and the various other agencies with an interest in this field as well.

The flight-medicine laboratories of the United States have pioneered in almost every area of bioastronautics—including acceleration. If we have lagged in applying these pioneer studies to operational systems, two primary reasons can be cited.

One is the well-known fact that we have concentrated, until now, on relatively compact booster systems. They are more than adequate for their intended task, which is to deliver military payloads for long distances over the earth. But they have been less than adequate to carry human flyers, with their complex life-support equipment, on extended missions into space.

The other reason—which is not so well known—has been the scattering of life science programs around the country, in a diversity of research laboratories, under many different authorities, without any clearly defined coordination or guidance.

The first of these difficulties—the lack of suitable vehicles—now is being remedied by NASA. The Saturn prototype was flight-tested for the first time, successfully, at Cape Canaveral one month ago. The Nova, as a backup system, and the Apollo moon vehicle are coming along.

The second problem—that of divided responsibility in the life sciences—has just been solved by the Air Force, with the consolidation of the Aerospace Medical Division. Assuming that full use is made of these facilities—by all the agencies involved in the national space effort—I am confident that the United States can achieve supremacy in manned space flight within the next few years.

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REVIEW OF NASA IMPACT WORK AND PLANS

Alfred M. Mayo
National Aeronautics and Space Administration

Dr. Hardy, General Bedwell, gentlemen—I have been asked to give you a little information on the interest and activities of human impact acceleration to the National Aeronautics and Space Administration. To meet its statutory responsibilities, NASA's interest in human impact acceleration must include three broad areas of effort. The first of these is a search for knowledge from and in support of space exploration. This work must involve the use of lower forms of life as well as human beings. Man will be important as both a test subject and an observer. Many of the early experiments in the basic science area can be conducted using remote instrumentation. As soon as our requirements become a little more sophisticated, though, it appears likely that the need for high resolution of information will force the human being closer to the source of his information. As a result there will be studies in impact using both remotely controlled and directly operated systems involving human beings. In addition to this search for knowledge and support of space exploration, we have an important area of research and development to answer specific problems related to manned space flight.

The Apollo project is broken into three general missions (see Figure 1). The first of these is an earth-orbiting mission and it has two primary purposes. The first purpose is to extend the time of investigations on human capability initiated in Project Mercury. The second purpose is to learn how to operate effectively in space. These earth orbital operations mean a great deal more than just a single vehicle flying around the earth. It is from this work that we will get our initial information on rendezvous. One can readily imagine the importance of knowing the limitations of the man with respect to the impact acceleration and how to design and protect him.

I would like to point out that in this work there are two important divisions of emphasis which should be kept in mind in determining the relationship of your scientific work to the actual program. The first of these includes those stresses incident to "normal" operation. In the normal operation the most important aspect is that of maintaining the highest possible performance of the human being in the system. A man should be kept under load conditions and operational conditions in which he can think clearly. In these early space-exploration efforts, the ease with which man can assimilate information and compare it to knowledge he already has is probably critical to rapid acquisition of new knowledge. Therefore, the "normal" operation, from the standpoint of impact as well as other loads, is one in which we should strive for a comfortable, effective environment for the man. The second division of emphasis has to do with what happens when things go bad. Here we have quite a different side of the picture, one in which John Stapp has done a great deal of pioneering. The problem is to discover what the man can tolerate in order to save his life. In this instance we have no holds barred, so to speak, in exploring the maximum capabilities of saving a man. These two conditions are quite different and as a result require quite different philosophies and approaches.

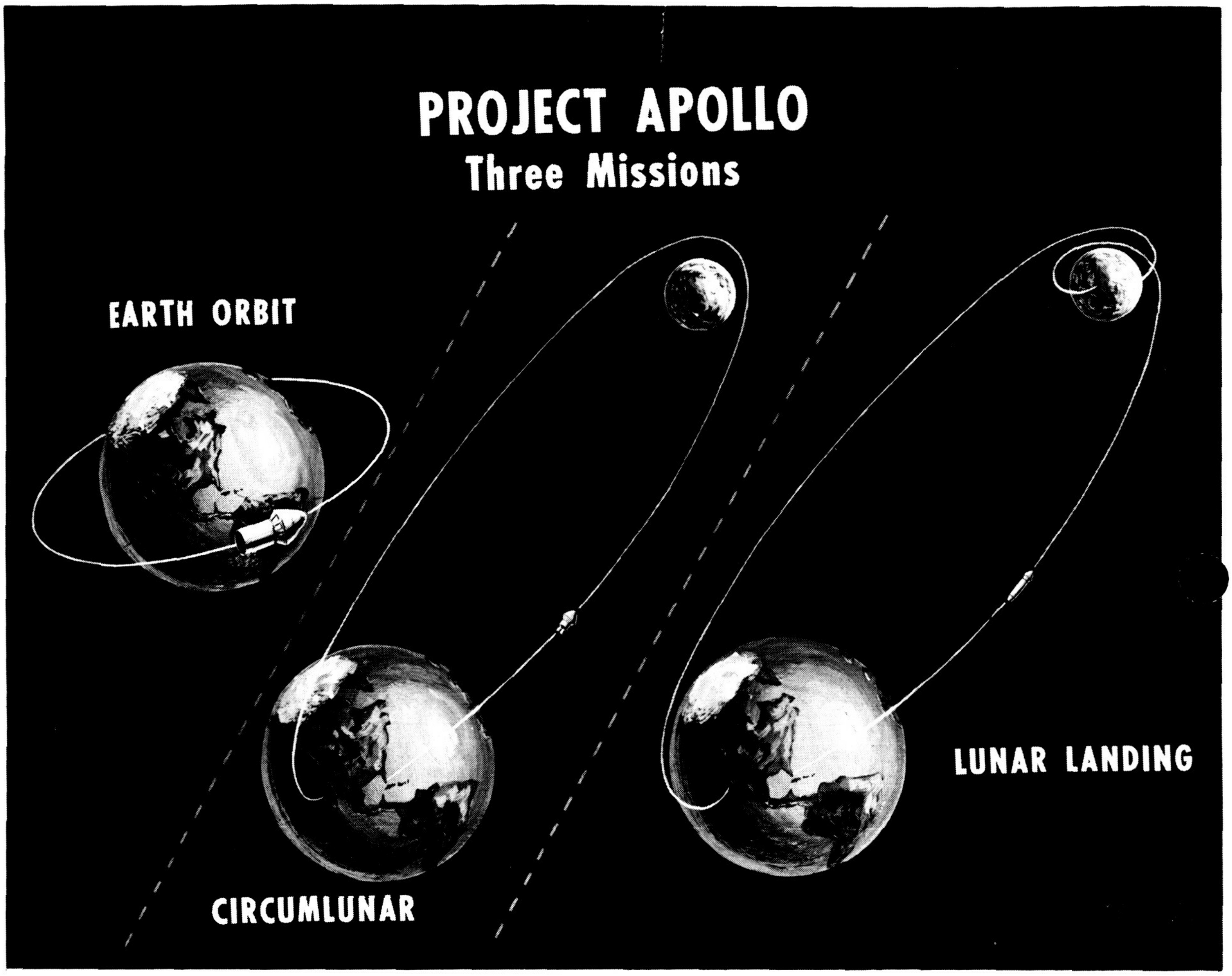

Figure 1

Imagine a mission in which vehicles are to be assembled in orbit. This operation may take place in an environment in which a man may not be sure of his orientation. In the early phases we may have instrumentation that isn't working as well as we would like to have it working. There will be situations occurring in which loads may be applied from almost any direction. From past work we have a fairly good envelope of information relative to tolerance along the major body axes. In this condition impact loads may be applied from any direction, with resultant combined loading. It is likely that we have a good deal more to learn about the potential effect of these impacts on human performance as well as on human structural tolerance. I believe our earth-orbital program particularly needs your help on these problems.

On the circumlunar mission we do not expect a great deal more in the way of impact problems; the impact incidents we might envision during the circumlunar missions would be no different from those in earth orbit. The kinds of major accidents envisioned are not too likely to be those that we can do a great deal about. It would appear that, of the missions shaping up, the one with the least relation to your work is the circumlunar mission.

When we come up to the actual lunar landing and return, we have the rendezvous problem extended from one in orbit to one in which we have to land in a vacuum on a hard object. During this mission, as you will see from the vehicles, we could have quite a number of load inputs depending on variations in normal and emergency operations.

At the moment we have three types of launch vehicles which are in a state of evolution (Figure 2). The evolution is particularly important with respect to the C-2. In

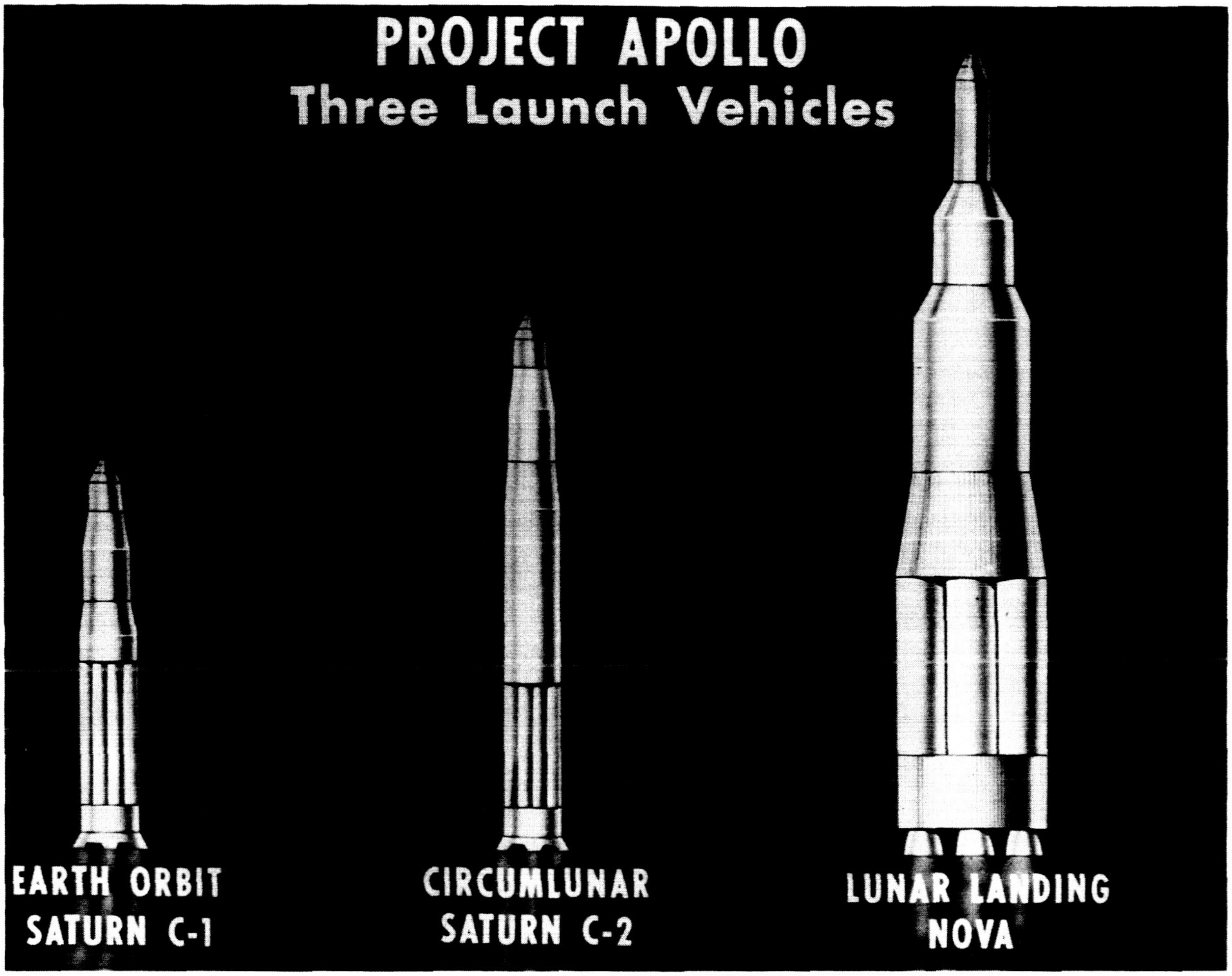

Figure 2

fact, it may not be called a C-2 at all; it may be called a C-3 or C-4 or some other number, but generally speaking the vehicle which flew last month, the Saturn C-1 vehicle, is the basic unit. The Saturn C-1 is adequate for a wide range of orbital missions with payloads which are high enough so that we no longer have the severe restrictions of which General Bedwell spoke. With the advance versions of this vehicle, depending on the final configuration of flight systems, we could do a circumlunar mission. The landing, of course, is something quite different, and as you well know, NASA is considering two approaches to lunar landing: (1) the lunar-landing Nova or the direct approach; and (2) the build-up of sufficient equipment in orbit.

Generally speaking we will have three kinds of spacecraft involved in this operation. One of those, with a relatively large mission module for doing extra work in earth orbit is shown in Figure 3. This system can perform a wide variety of laboratory

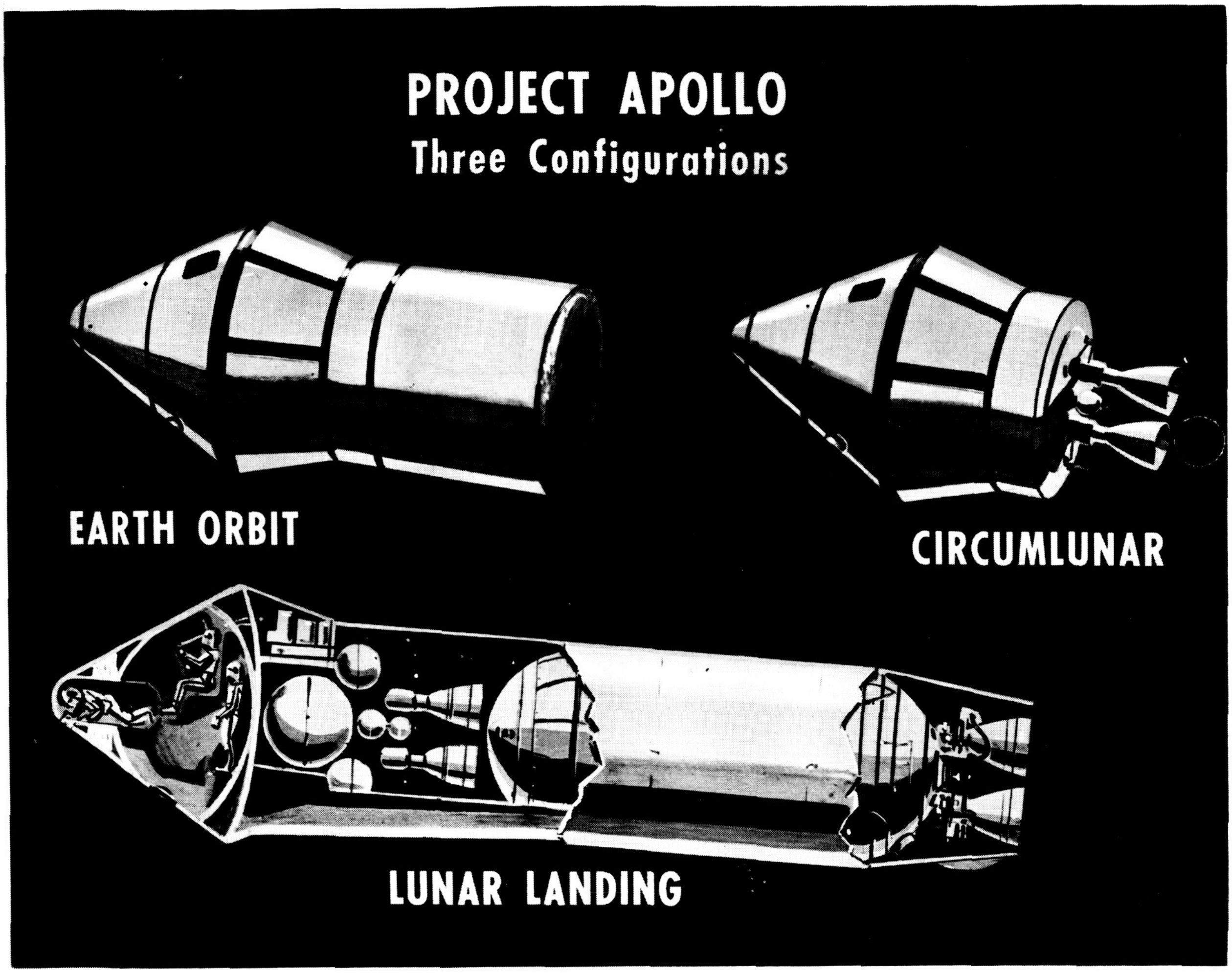

Figure 3

tasks, many of which should be of interest to the acceleration community. In the circumlunar mission, some of the laboratory functions have been given up for more propulsion in order to get a close look at the moon and to gather the information necessary for the follow-on landing. In this phase I would say the impact work would have its least immediate requirements. When you go to the next stage, the actual lunar-landing and take-off vehicle, one can readily imagine a whole host of new problems. If you will consider the artist's conception of one configuration (Figure 3), you can see that we are potentially interested in impact loads from a variety of directions and not just from the terminal direction. Of course, the decision has not yet been made as to whether this vehicle will land in the position shown, or will land on its tail. Much work is required before these final decisions are made. On the other hand, we do have, potentially, the whole cross-section of problems which have been previously studied lumped into a single project. I think that it will provide an immediate challenge and generate new ideas for future basic work.

Let us just go through briefly this lunar mission (Figure 4). First, let us consider the direct approach, in which we take off with the large Nova vehicle. After

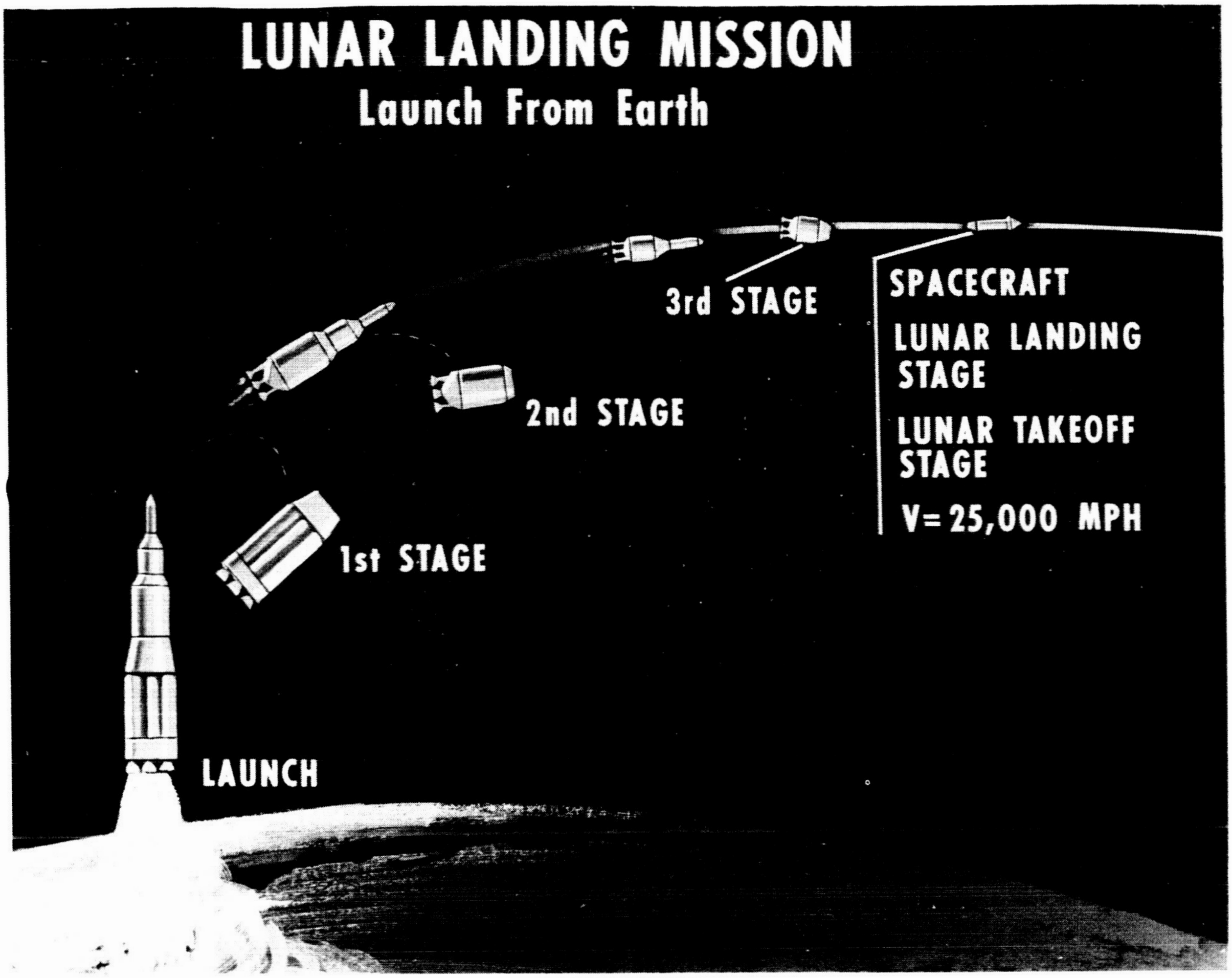

Figure 4

jettison of the first stage a large second-stage unit propels the system to orbital speed, and then an additional impulse provides escape velocity. The impact problems we start worrying about are concerned with the little man-containing thing on the top right. At takeoff there are a number of kinds of impact problems relating to such possibilities as catastrophic explosion, in which we have not only to consider direct mechanical-impact-type loading but also to worry about shock waves creating impact. There is also the possibility of vibration problems which could be serious. It is necessary to decide on a distinction between what we call an impact problem and what we call a vibration problem. Takeoff is a particularly critical stage when one considers the escape-sequencing, the firing of escape systems, and the landing that would occur in case of ground abort. As we move on into the orbital stage of the operation, in the direct approach, we would not expect to have too many impact considerations. In the orbital-rendezvous approach, on the other hand, we could encounter a number of impact problems.

To come down onto the surface of the moon we have a retro-thrust system (Figure 5). Under optimum and normal conditions there will be a nice smooth gentle

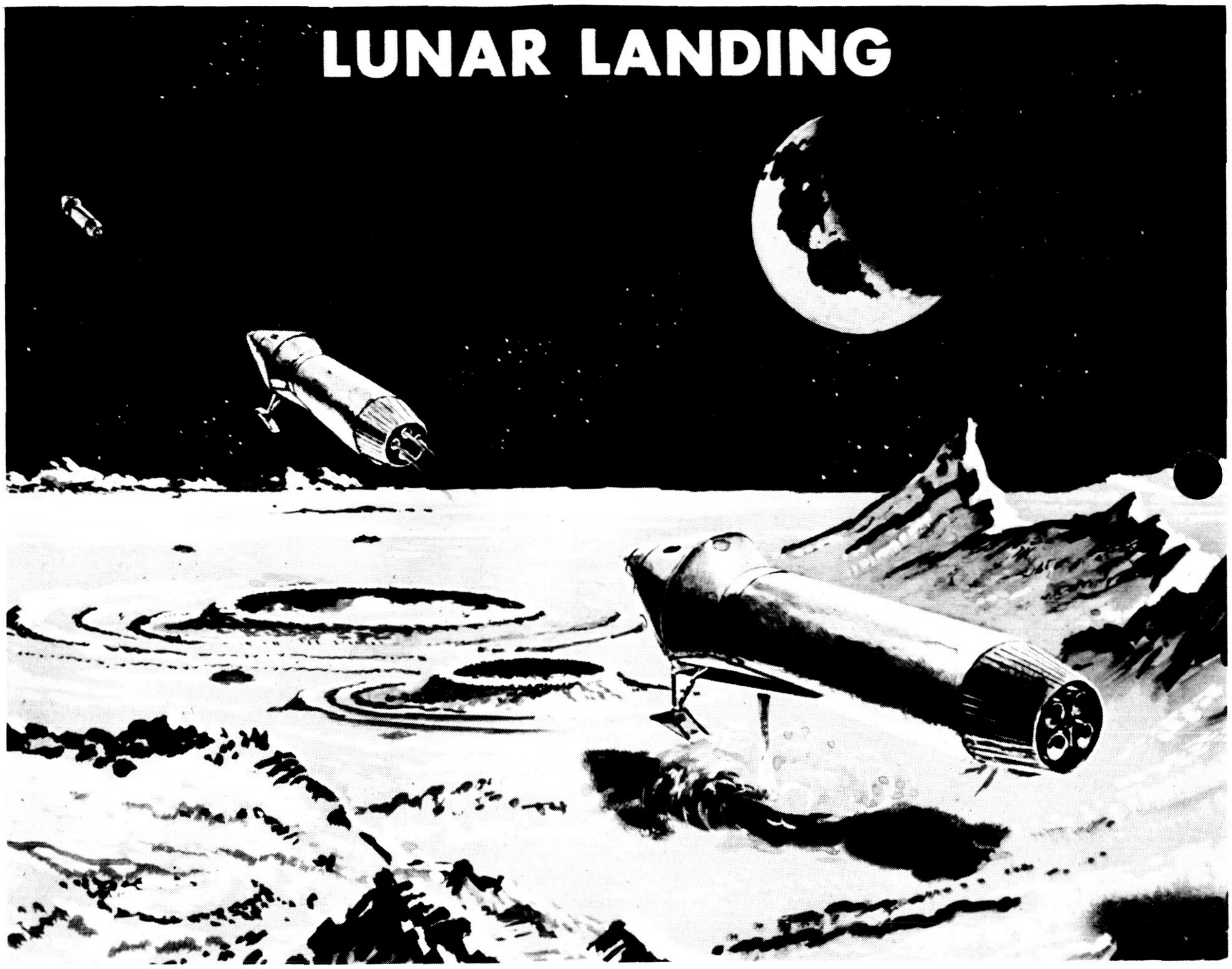

Figure 5

landing. With a suitable landing gear, the impact will probably be no more severe than the landing of an aircraft. On the other hand, since we are working in a frictionless environment and since we will not have had a great deal of experience with this type of device, one must take into account the possibilities of various malfunctions which would raise a whole host of potential problems for which you people could provide answers. These problems will relate both to the positioning of crew members and to means of restraint which will keep them in a completely operational state after landing. If the landing in any way injures the crew, you can see the chance of getting back is very slim indeed. This operation is highly critical and will be most interesting. Careful thinking in advance of the operation will be important to the proper identification of research needs.

The lunar takeoff will also be of critical importance with this type of system. Figure 6 shows how you might take off from the surface of the moon, leaving the landing vehicle on the moon and taking off with only the earth-return module and the atmospheric-reentry module.

Figure 6

As we get back to the atmosphere, of course, (Figure 7) we are coming back to a host of problems which have to do not only with the smooth accelerations associated with normal atmospheric return but also potential impact problems in the event of instabilities. Of course, in the final landing we have the same type of impact considerations that were important to the abort and orbital-landing parts of the mission. There are two separate classes of problems: (1) those of providing an adequate normal environment in which the man is comfortable and effective; and (2) those of providing optimum protection to maximize survival chances in an ultimate load condition.

As all of you people know, NASA's direct capabilities in the life sciences are small compared to those existing in universities, in industry, and in the military services. To do its job NASA must depend on the capability existing all over the nation to help get the job done. On the other hand, some life sciences work will be very closely involved with specific missions. Therefore, suitable means must be provided to ensure that the effort is one which will bear directly on the manned space-exploration problem in the early stages. At the same time it is important that knowledge be generated which can be applied also to the future needs of other agencies. A major area of important work in which NASA has an interest, but which it will pursue only indirectly, is the stimulation of industrial and military application of information from past research. In this connection I would like to bring up what I believe to be one of

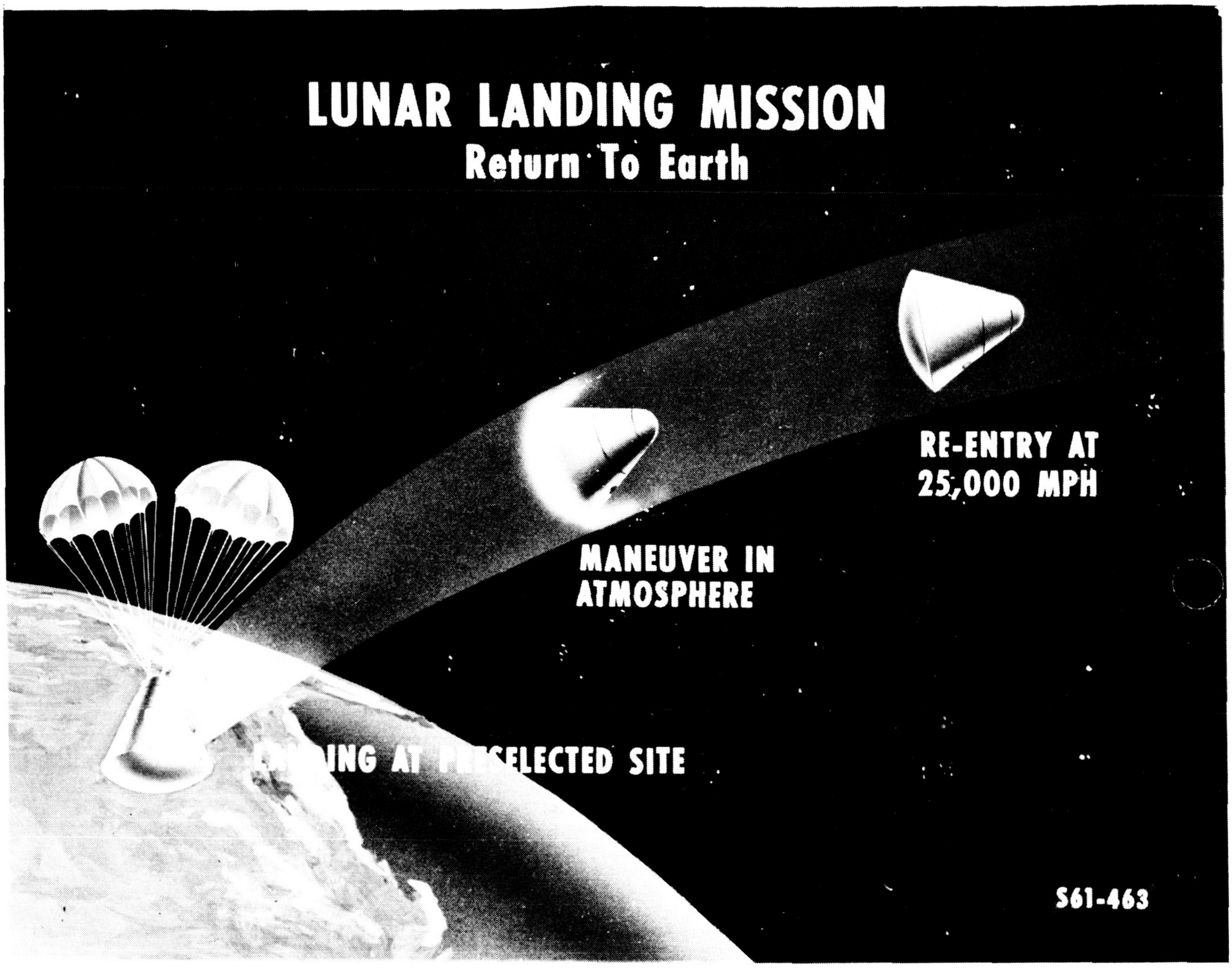

Figure 7

the most pressing problems in impact work at the moment. Very good information is available from past work—information which could be applied to significant improvement in the utility and safety of certain classes of military and private aircraft. Some improvements, of which I believe everyone is aware, have been made. The potentials of the information now available are such that a very important area of development is involved in making full use of it. In this respect NASA stands ready to cooperate with other government agencies and with industry to provide information helpful to stimulate the earliest practical realization of these gains. The responsibility for providing the needed development and facilities to apply the existing knowledge rests primarily with industry and with Government agencies outside NASA. NASA is now sponsoring research and is prepared to sponsor or conduct additional research and development which can result in improvements for aircraft as well as for its primary mission in manned space flight.

As I pointed out before, we can immediately see needs of limited-impact research to meet the orbital-rendezvous, the lunar-landing and the earth-landing requirements of the presently defined space mission. This research need includes the determining of human-tolerance impact along body axes, other than the normal horizontal, vertical, and transverse. There are other areas of acceleration which may or may not be called impact under certain definitions, in which it may be possible to extend considerably

the capability of the human being. Dr. Hardy, Dr. Clark, and others have long believed that we have not come to the end of the road in what we can do to protect man. I would like to urge that we take real cognizance of the possibility of jump gains in human capability to withstand acceleration by the proper loading of the human body and the utilization of protective devices, rather than expending all our efforts on making the marginal gains. This is the difference between what bold advanced-thinking research can do for us and the continual effort of inching along. Both approaches are essential, but I believe we have with us here the group that possibly can pull us out of the rough, so to speak, and cause us to think toward larger gains. I believe it likely that these larger gains would be mostly in the direction of improved survival when things go wrong, rather than of further definition for the normal state, at least with respect to the impact loading.

The National Aeronautics and Space Administration solicits the advice of this outstanding National Academy of Sciences panel of acceleration experts to help identify and promote a dynamic and effective program. As I pointed out before, the major capability in these areas lies outside of NASA. According to the law, NASA has the primary immediate responsibility in the United States for conducting the manned space-exploration programs. The answer is very simple; we must align the capabilities that exist with the responsibility in NASA to get the work done. In the area of acceleration, and particularly today in impact acceleration, we have a cross-section of the world's most capable people here. Thus, I would expect that the advice we can get from you people will be the kind that can promote a rapid increase in progress and knowledge in impact-acceleration. Thank you.

Discussion

Dr. Hardy: Question. How can NASA's work requirements be met?

Mr. Mayo: Well, real simply, the plan that is being pushed most at the moment is to find means of utilizing the capabilities that now exist wherever they are, and at the same time, in the light of the urgent needs, to expand these capabilities. I am quite sure that these expansions will occur in many places. The expansion you see out in front of this Center is just one example. We are also drawing capabilities from very important work in the Navy installations. We are drawing important capabilities from universities and industry. I think it would be completely out of line to assume that any one of these alone represents the capability of the United States. Success depends upon finding a means of directly utilizing these available groups. Now there has been a start, of course, in the university area under grants and contracts work, and there is a start in terms of military and NASA making organizational changes to work together more easily. All of the detailed answers are not yet available, but I would expect that you will see significant progress in this direction in the next few months.

Dr. Hardy: Question. Do you believe that we have done all we can to use presently available information to improve automobile safety?

Mr. Mayo: I believe that this is an excellent example of the point I was trying to make about ways in which we can make very great use of the information already available while new information is being gathered, and, of course, automobiles being the greatest in number, can be one of the beneficiaries; but

I can well imagine that Dr. Goddard's light aircraft will be increasing in numbers and I know that the military will have important small-aircraft operation. I really believe that we have need for this group to identify testing programs and development programs to ensure that its past information is applied, as well as to identify areas for future work.

Dr. Hardy: Is there a real problem here in that we have a lot of information lying around that folks don't know about?

Mr. Mayo: I am not so sure that it is a matter of information that folks don't know about. I think it is a matter of active stimulation of the development and use of that information. Perhaps the fertile minds of the people who help produce the information when rubbed up against some of the practical problems may also produce some of the design approaches. They may also help define some of the testing and development facilities necessary to make better use of the information now available.

Dr. Hardy: Question. Is all of the available information in a distinctly usable form and is it being made completely available?

Mr. Mayo: Well, there is some help possible in this direction. Because of the need of both NASA and the military for ready access to information of this sort, there are moves underfoot again to provide funds for the sort of thing to which you refer. I think that with adequate coordination of effort this information can be put into such systems as the ASTIA system and the NASA general information system. I think this committee could help to identify for these indexing groups the kind of information that should be included. I would like to point out one other thing that has occurred; that is, the cooperative work of certain of the physical and life science groups. Payne and Dr. Lombard, for example, have participated in a number of efforts of this sort. They have combined knowledge of physics with medical acceleration work. The output has been usable mathematical quantitative relationships. One of the great weaknesses in the aerospace medical effort has been the lack of quantifiable data of the kind commonly applied in the physical sciences. Now, when I speak of quantifiable data I am not in any way implying that we do not have to take into account the statistical range of variations. We have to do that in engineering material; the precision of quantifying them is nonetheless important. I believe that as the available data are put into a form in which one can weigh the medical versus the physical consequences in a certain course of action, we will be in a much better position to ensure appropriate consideration of the medical aspects of a problem or of the physiological aspects of acceleration problems.

There is one request I have for you people in this particular area. Put yourself in the position of us bureaucrats in trying to justify an expenditure for certain facilities and certain capabilities to do work. You can see that it would be valuable if we had from the scientific community, wanting to do that work, a better spectral picture of where a particular new facility fits into the range of those needed. Such a picture could make it easier for us to show that the new requirement is really expanding our capability and not just duplicating an existing capability. Also we need to know

where there is an honest need for duplication because of the size of our country, to allow people to interact with facilities. You can help also by identifying the importance of this need and by showing that the scope of the work is such that you may need two "lathes" to do a job, instead of one.

Question: What do you refer to when you speak of protection against explosions?

Answer: I referred to what could well happen in various types of explosion with hydrogen, oxygen, and other types of fuels that are used. These gaseous shock waves themselves could apply both mechanical impulses of the normal type and impulses which may only be a single cycle but border on the sound and vibration type, in which we are working at high energy levels and low-repetition numbers.

Question: Will these explosions apply loads to the capsule or to the man?

Mr. Mayo: Obviously these phenomena can cause both; you can have the pressure forces applied to the whole vehicle, which will give significant shock in the direction of which the major force is applied, and you can have also a shock wave which travels through the vehicle and applies forces directly to the man. You need to identify these ranges and provide for suitable attenuation in either case. It may be that information already available will show that one of these is unimportant. On the other hand, it is desirable to have the information in such a form that people utilizing it in the design of space vehicles do not waste their time, but are able to concentrate their efforts on those critical problem aspects. Does that answer the question?

Question: Will the vehicle itself not give the man some protection against these explosions?

Mr. Mayo: That is right, and I think, incidentally, there is evidence to indicate that some of these added safety potentials may well be there when you consider the really tough capsule that is created when reentry material and protective material to the thickness of several inches are placed around the small vehicle. It may be a lot stronger than most people believe.

12546

REVIEW AND FORECAST OF IMPACT STUDIES
UNITED STATES ARMY

Major James C. Beyer
Office of Surgeon General, U. S. Army
Washington, D. C.
and
Captain Anthony A. Bezreh
U. S. Army Board for Aircraft
Fort Rucker, Alabama

The aviation program in the United States Army is a relatively new effort; thus, there is no very extensive set of data on impact experience, the analysis of which advances the frontiers of knowledge on this subject to any great extent. Moreover, extensive programs or facilities for investigation have not so far been established.

However, because the problems of impact stress are especially pertinent in Army flying, the need for study of these problems has been recognized, and significant research efforts are being launched.

In order to understand the Army's stake in this problem and the kind of information which can be gained, one must understand the mission of Army aviation and the aircraft with which this mission is to be performed. The mission of Army aviation is to provide the ground forces with the mobility and intelligence demanded by modern warfare. The type of flying required involves, for the most part, fairly low altitudes and relatively slow air speeds. The aircraft best suited to the combat conditions in which Army aviation operates are those which can be used on unimproved strips and have short takeoff and landing capabilities. These are light aircraft, more than 50 per cent of which are rotary-wing. This preponderance of the rotary-wing types is expected to become much more marked within the next few years.

Thus, it is self-evident that the Army must have as keen an interest in impact studies as do the other services. The Army pilot, when confronted with an accident situation, cannot eject, because most Army aircraft are not equipped with ejection seats. He usually cannot bail out in the conventional manner because of insufficient altitude. If he is flying a helicopter, he does not even have a parachute at his disposal and must go down with the machine. In any case, the Army aviator usually has to "ride his aircraft in."

The saving grace of the situation is that he most often "rides his aircraft in" at a relatively slow airspeed, and the crash which results is usually classifiable as survivable. In other words, the crash forces generated fall within the limits of human tolerance, and the amount of destruction of the inhabitable area within the aircraft is not so great as to impinge upon the vital organs.

Although the line between accidents which are classified as survivable and those classified as non-survivable is sometimes difficult to draw, and since this is often a

subjective decision, an analysis of 1,200 major Army aircraft accident reports conducted at the United States Army Board for Aviation Accident Research, indicated that an impressive 97 per cent of Army aircraft accidents may be considered as survivable, whereas only three per cent of them are non-survivable. Furthermore, the happy prognosis of survivability applies equally well to both fixed-wing and rotary-wing type aircraft.

Of course, when the definition of survivability depends on the limits of human tolerance to crash forces, and when these limits of human tolerance may be considered still essentially unestablished and open to experiment, one must regard figures like these with critical reservations. The further difficulty of computing the G-forces involved in an aircraft accident with any degree of accuracy does not help in the assessment of survivability.

However, in spite of these problems, it can confidently be said that by far the majority of Army aircraft accidents are survivable. This is in sharp contrast to the accidents experienced by the other military services, in which high-speed aircraft crash, with resulting total demolition of both man and machine. Therefore, the opportunity and responsibility exist in Army aviation to gain valuable information on how to design crashworthiness into aircraft, how to develop improved protective personal equipment, how to incorporate effective restraining and energy-absorbing devices, and, thus, how to prevent crash injuries and fatalities.

Theoretically, there should be no fatalities in survivable accidents. However, even with the tolerable crash forces being experienced in Army aircraft accidents, the Army is far from attaining this ideal goal, because of deficiencies in aircraft design and in the use of protective equipment. It must be stated, with proper remorse, that as many Army pilots have been killed in survivable accidents as in non-survivable ones.

One of the major facilities which the Army has for coping with such aeromedical problems is the United States Army Board for Aviation Accident Research. The contribution of this organization on the impact-deceleration problem is made through an empirical approach involving analysis of Army aviation-accident reports and first-hand crash-injury investigations.

Naturally, the first step was to determine the nature and relative importance of impact-deceleration injuries with respect to the complete crash-injury picture. When the injuries which can result from impact decelerations are being studied, injuries produced by the effects of forces alone, as determined by the structures and strengths of human tissues, must be clearly differentiated from those produced when the machine and the man are thrown against each other.

An analysis of crash-injury data from survivable accidents revealed that only six per cent of the injuries incurred in this category of accidents could be attributed solely to the pure effects of forces on the man. The reasons for this small figure are that (1) many of these accidents do not generate forces of sufficient magnitude to exceed the strengths of human tissues, and (2) perhaps more importantly, the number of injuries produced by pure force effects is overshadowed by the much greater number of injuries produced by direct man-machine contact.

To look briefly at the injuries produced by pure decelerative forces, and being careful to avoid making unjustifiable generalizations from a relatively small body of

data, it is apparent that the great majority of these injuries, specifically about 70 per cent, involve the spinal column. These spinal-column injuries have occurred uniformly throughout the cervical, thoracic, and lumbar regions, and have ranged through compression fractures, dislocations, and muscle strains. Perhaps the one notable thing about these spinal-column injuries is that 75 per cent of them have occurred in rotary-wing accidents in which vertical decelerative forces have been dominant.

In view of the fact, then, that an increasing proportion of the Army aircraft inventory will be of the rotary-wing type, and because the newer models of these helicopters exhibit comparatively higher sink rates, which, of course, increases the frequency and magnitude of vertical decelerative forces during the crash sequence, it may be expected that Army concern and research may eventually be directed particularly toward the effects of vertical decelerative forces and the means by which they may be attenuated.

Most of the remaining injuries produced by pure decelerative forces involve the rupture of the various soft-tissue organs, such as the heart, possibly through the common denominator of the hydrostatic effects resulting when the erectly seated aircraft occupant is exposed to vertical decelerative forces, or when the forward-flexed occupant is exposed to longitudinal decelerative forces.

Obviously, man's tolerance for these forces is directly related to the adequacy of his tie-down chain, including the shoulder harness, seat belt, and seat, and to the energy-absorbing capabilities incorporated within this tie-down chain. It is toward this tie-down chain that much of USABAAR's effort has been devoted; for, in almost every instance, this restraint equipment presently available in Army aircraft cannot withstand the forces that can be tolerated by the human body.

To compound this problem, very few Army aircraft have cockpits that can withstand crash forces that even approach the most conservative estimates of human tolerance. A great deal of USABAAR's aeromedical effort has been devoted to improving this situation also, because, as long as these conditions exist, it is almost academic to be concerned about the relatively minor role of the effects of pure decelerative forces in injury causation. However, it is hoped that these conditions will not continue to exist, as attention to considerations of crashworthiness and of the restraint system is gradually increasing and features are gradually being incorporated into the newer aircraft which will bring the crash-endurability of hardware more into line with the crash survivability of the "software," or human occupant.

It should not be concluded that the less-than-ideal crash protection provided by most Army aircraft results from neglect of the problem, but it is no easy task to put a 40-G cockpit into a light observation helicopter and still have the aircraft able to get off the ground. Probably even the most conscientious attempts to provide crash protection in light aircraft will have to be compromised by such practical considerations.

Thus, because it has been necessary to combat the more elementary problem of simple traumatic impact injuries, relatively little attention has been paid to the somewhat more sophisticated problem of pure decelerative injuries. Naturally, as the former problem is gradually solved, the latter will assume increasing importance to the Army, because the adequately restrained aviator in a crash-resistant cockpit will then suffer no injuries, other than those resulting from pure force effects.

In an attempt to establish exactly those forces which have to be dealt with in light-aircraft accidents, the Army's Transportation Research Command established a contract with Aviation Crash Injury Research, a division of the Flight Safety Foundation, to conduct carefully monitored, instrumented drop tests. In this program, light aircraft are being purposely crashed under control conditions, and the resulting forces are being quantitatively and qualitatively measured with respect to magnitude, direction, frequency, rate of onset, and such other variables as are known to be of importance in the production of crash injuries, in the development of protective equipment, and in the design of crashworthy aircraft structures.

The method of this approach to the problems of decelerative forces has been gratifyingly successful and, although these studies have been started only relatively recently, a great deal of valuable information has already been obtained; much more can be expected to follow. The approach in these tests is to define the problem in a truly scientific manner, avoiding the guesswork and approximations inherent in after-the-fact approaches, and to test possible answers to the problems at the same time.

As a result of the recognition that the missions and machines in Army aviation pose certain aeromedical problems of crucial interest to the Army alone, it has been decided to establish a United States Army Aeromedical Research Unit, which may eventually contribute to the Army's efforts to evaluate and attenuate impact-acceleration stress. An example of the type of problem peculiar to the Army aviation mission is that of the effects of low-level turbulence, with its various amplitudes and frequencies, experienced by the pilot of a relatively high-speed reconnaissance aircraft flying at altitudes of 200 feet or less in order to avoid radar detection. The effects of a rapid succession of positive and negative jolts for extended periods of time have had to be studied to ensure that a pilot would not be physiologically or psychologically incapacitated for such a difficult type of flying.

Finally, it must never be concluded that solutions to a problem must come only from formal programs and facilities aimed at and dedicated to the particular problem or problem area. Answers are to be found wherever there are men with imagination and ingenuity. For example, two chemists working on plastic foams for the Army Quartermaster Research and Engineering Command have come up with a proposal that it is completely feasible to develop a crash-activated system which would contain the chemical components of an extremely rapidly forming and highly energy-absorbent plastic foam. The system, when activated either by the forces of a crash sequence or by a panic button which a pilot could elect to use if he felt that a crash was inevitable, could spread a plastic foam in some optimal amount over the injury-producing surfaces of the cockpit, which would then harden rapidly enough to attenuate survivable decelerative forces and undoubtedly reduce a certain range of non-survivable decelerative forces to a tolerable magnitude.

This necessarily has been a brief survey of what have been and will be the Army's studies on the problems of impact-acceleration stress. What is learned from these studies will not only be meaningful in terms of Army mission effectiveness, but also will undoubtedly be of benefit to civil aviation, as a bonus, because of the overall similarity between light civilian aircraft and Army aircraft.

12847

FEDERAL AVIATION AGENCY IMPACT RESEARCH

James L. Goddard
Federal Aviation Agency

We are delighted to have the opportunity to discuss the research activities of the Federal Aviation Agency in the field of impact acceleration stress. This highly significant field is of increasing interest to civil aviation. The steady growth of air carrier and general aviation activities is inevitably accompanied by various types of accidents.

Of particular concern to us in this regard, is the "severe-but-survivable" accident. We know that the sudden deceleration forces in these circumstances can be dissipated in such a fashion that the passengers and crew experience only non-fatal injuries. The basic structure of the airplane, its flight equipment, the seats and seat belts, play key roles in protecting the occupants of a crashing plane.

The Aviation Medical Service, through its Division of Research Requirements, is supporting studies of the above factors, combined with studies of the physical characteristics of the victims in relation to their injuries. Our Civil Aeromedical Research Institute in Oklahoma City, a component of the Division of Research Requirements, is focusing attention, through the Protection and Survival Branch, on the physical interplay between man and machine during sudden decelerations.

The Aviation Medical Service's responsibilities include the matter of emphasizing "crash safety" through developing and making available information on crashworthy aircraft structure. Our goal is the elimination of needless loss of life as a result of non-crashworthy structures in aircraft. We are cognizant of the several specialized studies in the broad field of impact acceleration stress, conducted by the National Aeronautics and Space Administration, the Department of Defense, the Flight Safety Foundation, and certain other organizations.

Our studies are oriented toward the conditions characteristic of civil aviation. The following illustrations describe certain of our activities in this vital field.

In general, our research can be listed in three major categories:

1. The further determination of human tolerance to impact forces;

2. The further determination of forces transmitted to the occupants of current civil aircraft during crash impacts; and

3. The further development of crash safety designs which can contain the crash forces transmitted to aircraft occupants at levels below the maximum human tolerance level.

Since our primary concentration of combined biomedical-engineering talent exists at the Civil Aeromedical Research Institute, the bulk of the studies which follow,

fall immediately within its purview. Many of the studies are currently in progress at the Institute.

In regard to the human tolerance to impact forces, we have completed and published the results of a series of laboratory vertical impact studies. We are currently planning a continuation of these studies, which will incorporate a greater degree of medical support. Specially designed vertical and horizontal decelerators will be used to simulate forces which might result in actual crash situations.

Dynamic and static testing of various body segments of cadavers is presently under way. For example, we are collecting information on the magnitude of forces produced and the location of failure points in the lower leg in relation to seat tubes. We are undertaking an extensive program on this particular matter.

In selected aircraft accidents, we are obtaining detailed medical histories and autopsy reports, which aid in relating injuries to aircraft structure.

We are also studying non-aircraft human impact acceleration injuries at a rate of five hundred per month, to round out the fund of knowledge relative to tolerances under different conditions of impact and acceleration. Gaps in this study will be filled in through certain animal studies.

We have published a study of the center of gravity of the individual human being. The center of gravity determines the nature of the kinematics of the body during certain impacts when restrained by a safety belt. Of interest here, in view of the wide variation in age groupings found among airline passengers, are our studies of the center of gravity of infants and children. Extensions of this study will involve compressibility determinations of human tissues and frequencies of oscillations of different organs in humans secondary to impact.

With respect to the actual forces which occur during an aircraft crash, we intend to make certain highly refined measurements along these lines on two specially designed acceleration tracks. Information to augment these studies may be derived in the future through the placement of certain recorders in commercial aircraft. In the event of an accident, the recorder, currently in the conceptual stage, would be salvaged and would yield information on the magnitudes and directions of the forces involved.

We have a theoretical study in its initial phase which involves investigating the energy absorption characteristics of various aircraft structural configurations. Also, we plan to flight test light aircraft for the purpose of obtaining information on the probable impact velocities, angles, and attitudes which are associated with accidents in airplanes of this type.

Studies on seats and seat tie-downs in relation to impacts of different magnitudes are planned.

With respect to studies of methods of reducing the forces transmitted to aircraft occupants, we plan to study certain energy absorbers in seat structures and tie-downs. We are seeking better means for the delethalization of the area surrounding passengers and crew.

Field investigations of crashes involving new model light aircraft are underway. Recommendations to improve crash protection for occupants are expected from the studies during the early operational period of these aircraft.

In conclusion, I should like to observe that about ninety percent of the total impact acceleration research effort in the United States today is concerned with space capsule re-entry and impact problems. The Federal Aviation Agency, on the other hand, is focusing its efforts on impact problems in civil aviation. Undoubtedly, certain of our findings will be of value in these other pursuits, and vice versa. We will, therefore, continue to keep in close touch with all organizations conducting research on impact acceleration stress.

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BIOMECHANICAL PROBLEMS OF THE LUMBAR SPINE

Olof Perey
Orebro, Sweden

If a man drops down from a relatively low height, landing on his feet with legs straight, he easily gets a fracture of the os calci. It is a rule always to examine the back, because this fracture is often combined with fracture of vertebrae. If both os calcii are fractured there should be an x-ray examination of the spinal column. On the other hand, if the man falls on his gluteal region and is unfortunate enough to get a fracture, it is a fracture of the vertebral column, except three per cent which are fractures of the pelvic ring. The vertebral column is from a mechanical point of view a very vulnerable part of the body. From a clinical point of view, we often hear patients declare that they feel something breaks in their back after a dynamic or static stress. The x-rays are negative; still something must have happened, but what?

To get an answer to this question I have done experimental investigations on specimens consisting of two vertebrae with the intervening disk, or three vertebrae with the intervening disks, from a great number of individuals. I have used both static and dynamic stress. The inner part of the disk has been visualized by diskography. The course was followed with x-ray films with 48 pictures/sec and with pressure curves.

I found that there was a great biological variation on the breaking-point. It was statistically clear that the breaking-point on individuals over 60 years old was much lower than in younger groups. The average values in the group over 60 years old was 425 kp. In individuals under 40 years old the breaking-point varied between 500- and 1,100 kp, with an average value 790 kp. In one-third of the specimens I get fracture of the vertebral end-plates. These fractures can be divided into three different types:

a. fractures centrally in the end-plate

b. fracture of the end-plate situated so far peripherally that a corner of the vertebral body is torn loose

c. fissure extending across the entire end-plate which, when deepened to involve the whole vertebral body, divides it into two parts. This is the first stadium of the common compression fractures.

As an accessory finding I can mention that the osteophytes are very fragile and they fracture even for 100 to 150 kp. To get a better understanding of these fractures I have examined the resistance both on the whole vertebral body and on different parts of the end-plate. The resistance of the body varied between 400 and 1,100 kp. I have also analysed the disk from different points of view. The result is this:

In a newborn child, the nucleus is semigelatinous and contains 90 per cent solution. Normally it desiccates with age, and at 70 years of age the nucleus contains only

65 per cent solution. This is a normal change, depending on age, called degeneration. In a young individual with a high amount of liquid, the nucleus takes three-fourths of the weight on the disk. The weakest walls in this room are the end-plates. With increasing force the intradiskal pressure increases and finally the end-plate fractures. This is demonstrated both analytically and experimentally. You never get a hernia by stress from a disk in this condition. With a dryer disk the pressure is transferred more and more by the annulus, and when the breaking-point is reached, that part of the vertebral body which is involved by the Sharpeys fibres fractures. Fracture in the frontal part is common and easy to diagnose, but in the dorsal part it is difficult to demonstrate on the x-ray film because of the kidney shape of the vertebral body.

A usual question is: What is the real strength of the vertebral arches and the articular process? In the older literature they say that this is the strongest part of the vertebrae. Now we know that this part only takes 20 per cent of the weight.

We can say that mechanically the lumbar vertebrae can stand a static stress of 800 kp or a dynamic stress during 0.006 sec of 1,300 kp. These values are quite low and something else must be involved in the weightbearing.

If you take a look at gorillas, you will see that they are heavy and very strong. The stress on their lumbar vertebrae must be high. The diameter of their vertebral body is less than that of the human. Fifty-five Swedish airforcemen have been catapulted, and of those 13 have gotten fractures mainly of the type I have shown from my experiments. The acceleration during 0.05 sec is 22 g, which means about 800 kp. This is near the breaking-point, but only 25 per cent get fractures. This must depend on muscle power, which has a much larger significance than we had believed earlier. All of you who are sailing know how important the guy-rope is for the mast, and I am absolutely convinced that the muscles are of the same great significance for the vertebral column. Erector trunci keeps the back upright but the abdominal muscles take part in the weightbearing of the body. If you have a long, thin balloon and increase the pressure, it will become stiff; in the same way you can keep the body upright by increasing the intra-abdominal pressure. Bartelink, and later Morris, Lucas, and Bressler, have demonstrated through EMG and by measuring the intra-abdominal and intra-thoracal pressure how these structures contribute to weightbearing. These investigations are just starting, and I am sure that they will reveal several pieces in the great and difficult biomechanical puzzle.

Other research tasks which still are unsolved relate to the resistance of the ligaments. I especially think of the interspinos-ligament. Ruptures in this ligament arise from dynamic forces, and produce back pain. When it heals with connective tissue—this being a poor tissue—it easily breaks again.

An interesting problem is the rotation of the lumbar spine. The intervertebral joints admit movement in two planes, but they limit rotation in the highest degree; otherwise the disk would be destroyed by torsion. The segment of movement in the lumbar spine can be compared to a universal joint. If you put a series of such universal joints together and have the whole series in a line, you can't rotate, but if you bend the whole series into an S-shape you can get a good deal of rotation. That is why we have kyfosis and lordosis. But what happens when you don't have the lordosis?

In conclusion may I emphasize that the spine is, from a mechanical and clinical point of view, a very weak part of the body, and therefore we must try to solve every problem related to it.

12-849

THE PHYSIOLOGIC ACCELEROMETERS

Dr. M. P. Lansberg
National Aeromedical Center
The Netherlands

The acceleration that we are dealing with in otology during high G and zero G is mainly concerned with sense-organ physiology, i.e., with our inbuilt linear and angular accelerometers. That brings to mind, Mr. Chairman, that you mentioned this morning that you regretted that no Russian delegates were here, and I am sure that you had in mind at that moment the kind of experiences Mr. Titov could have told us. As you know, his experience during his trip around the world were not all of a pleasant kind and, perhaps, there might be an explanation purely on a vestibular mechanism which might also have some bearing on our present problem of impact acceleration.

Weightlessness will cause several difficulties, and it is by way of introducing an artificial weight that a solution for these difficulties has been sought. The artificial weight is brought about by making the space ship or space station rotate around its center of gravity, and Figure 1 gives an idea of the situation. It shows that the inhabitants are no longer floating, as was the case during weightlessness. They are now "centrifuged" toward the outside and have consequently found a centrifugal weight. That looks all very well, but there are some new, unexpected problems that the inhabitants are now faced with. This is illustrated by the queer movement that one of the inhabitants makes. It is a tilting reaction caused by the fact that the man has rotated his head around a vertical axis, and by so doing has inadvertently given an impulse to his semicircular canals. The magnitude of the impulse can easily be computed and will be up to 33 degrees per second in Wernher von Braun's concept of a space station. Whether this impulse will be a hindrance to the inhabitants' well-being and mental and physical proficiency cannot yet be answered for certain. However, the evidence is accumulating that it will surpass the limits of physiologic acceptability. Particularly the recent work of Graybiel's group in Pensacola has unequivocally demonstrated that the physiologic margins are much narrower than originally anticipated.

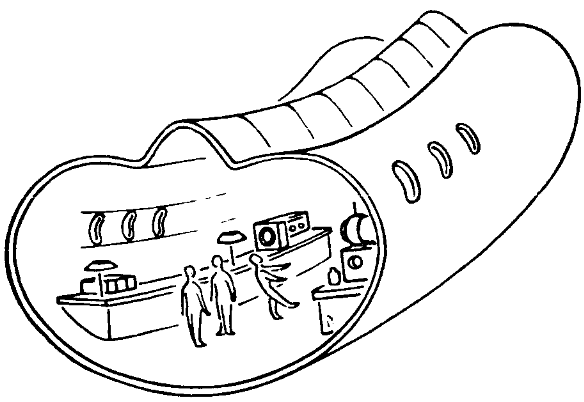

Figure 1. Life on board a rotating space station. Two members of the crew are discussing their problems; a third one just turned away and, turning around his longitudinal axis, is taken by surprise by "false" tilting sensation and tilting reaction.

The Pensacola experiments concern studies in the slow-rotating room, where the occupants are induced to move their heads every now and then around a horizontal

axis. This will elicit unexpected impulses in the semicircular canals, much the same as were just described for the rotating space station; as the slow-rotation room is found to be liable to produce unpleasant sensations and untoward reactions, the rotating space station will be no less provocative in this respect. Indeed, it will probably be still more difficult to endure. The reason is that it is not the impulse to the semicircular canals which causes the trouble—the impulse is not that large—but the conflict in signals that it evokes. In other words, everything would be easily tolerable if only the information coming from the semicircular canals, i.e., the tilting sensation, were not contradicted by the otolithic information which indicates that there is no change in perpendicularity. This contradiction upsets the occupant of the slow-rotation room as it will his colleague in the rotating space station. The situation is even worse in the space station, as we said, because not only head movements around a horizontal axis but also those around a vertical axis will induce "false" angular impulses. Whereas in the slow-rotation room adaptation to the "false" impulses was seen to occur in a few days, the more complicated situation in the space station might moreover well frustrate such physiologic tendencies. Even a simple forward bending of the head will for the space station's occupant be unpredictable in its effect, because the impulse to the canals depends upon the angle between the space station's axis and the axis around which the occupant is bending his head.

So much for the conflicts to be expected under artificial—centrifugal—weight. Let it be said in passing that this is still a theory, not adhered to by many students of motion sickness, who do not believe in the dominating role of the conflict between canals and otoliths or, more generally, between information coming from different sense organs in eliciting symptoms. They are still advocates either of a purely canal origin of motion sickness or else of a purely otolithic one. An argument against such a one-organ concept might be found in the following diagrams: The first diagram (Figure 2) is concerned with linear accelerations. In fact, the slope of the threshold curve in the lower region suggests that it is rather the first derivative of acceleration than acceleration itself to which the human body—*in casu* the otoliths—responds. Be that as it may, the upper curve, indicating the human tolerance to passive linear movements, is not surpassed under circumstances prevailing in traveling by rail or sailing (heaving of a ship). Whereas, under active movements like jumping rhythmically with a double amplitude of 20 cm, the otoliths will tolerate even considerably higher impulses. The second diagram (Figure 3) depicts the angular impulses that occur in different situations: rolling of a ship in rough weather, Barany chair as used in the clinical examination, the rotating space station, and finally a few examples of shaking the head as in a "No" gesture. These latter are included as physiologically occurring movements, and they will be seen to surpass by far the canal impulses sustained on a rolling ship or in the slow-rotating room or in the space station. Should not this argue again for the theory that it is the conflict in information rather than the impulse as such, that evokes the motion sickness?

Let us now turn from the artificial weight towards weightlessness. To cut a long story short, we meet once more an inherent contradiction (Figure 4). Three situations can be recognized.

a. The head is held in an inclined position relative to the vertical
b. The head is rotated around a horizontal axis
c. The head is moved along a horizontal path.

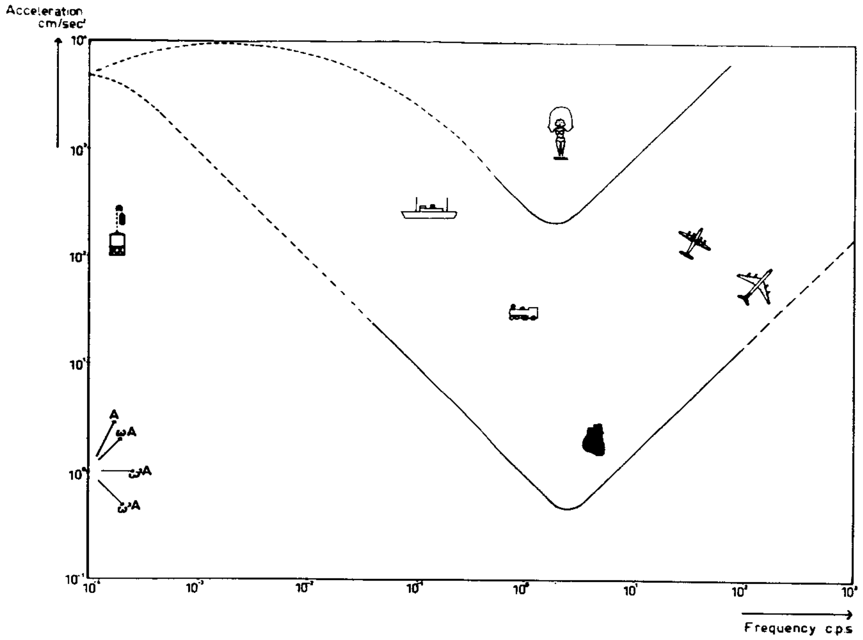

Figure 2. Magnitude and period of several linear accelerations.

The aeroplanes in this graph are located according to the vibratory characteristics of a Super Constellation and a DC-8, and not according to the movements induced by air turbulence.

The values for an elevator are 100 cm sec^{-2} for the normal one and 200 cm sec^{-2} for the express type.

Threshold is indicated by the lower curve. The limits, where passive movements become scarcely bearable, are given by the upper curve.

The slope of the curves in the region below one cps would suggest that it is the change of acceleration—ω^3A—which is decisive for the effects.

Active movements like springing or bending the head either sideward or forward can be quite well tolerated physiologically while executed in a pattern which imparts stimuli to the otoliths, surpassing those that were considered unbearable when accompanying passive movements.

The heart is used as a symbol for the oscillatory movements of the body induced by the heart's impulses as in ballistocardiography.

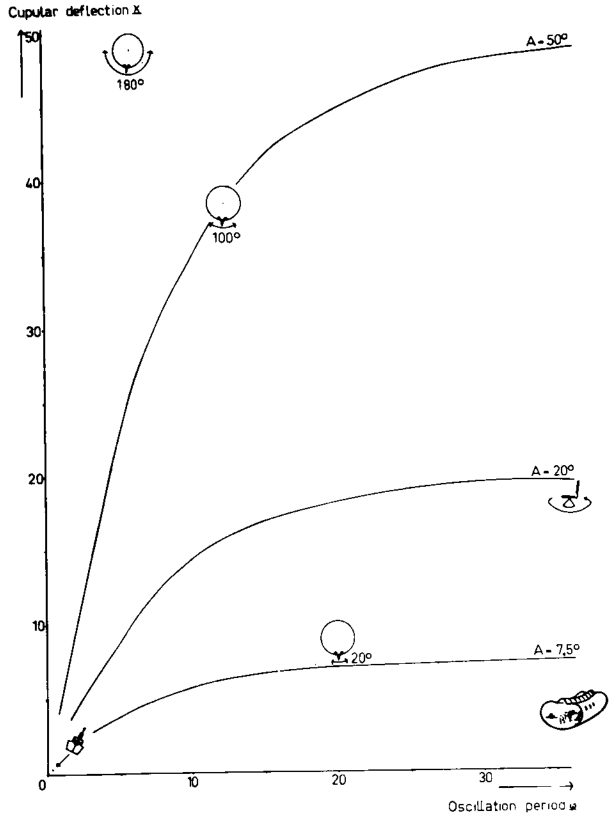

Figure 3. Angular impulses occurring in several situations.

The first question is what should be considered vertical or horizontal in a weightless situation. True, there is no graviceptive orientation from which these clues could be gathered. But man will of necessity shape his environment in an up-down fashion. During weightlessness it will be his visual framework that will induce his spatial orientation. Against this background the conflicting information may be seen to arise: In "a," the otoliths signal a symmetric status, albeit a symmetry of zero gravity, and for the higher centres a symmetry means perpendicularity, which is denied by the visual reference. Probably the voice of the otoliths will not be very strong in this instance, and the visual clue will dominate. The reason is that the otoliths, as said earli respond to change in acceleration rather than to acceleration.

In situation "b," things become worse. There is not only a conflict between otolithic information and visual framework but, worse still, the semicircular canals report a change in attitude which should be corroborated by the message from the otoliths which, however, makes default. Just as in the instance of the artificial weight, a disagreement now develops between the two parts of the labyrinth. The disagreement would not arise if the head movement had occurred around a vertical axis, vertical in the man's subjective framework of orientation.

Situation "c," will probably be of less importance because such linear movements do not seem likely to occur.

Discussion

Question: Dr. Lansberg, I am wondering why you consider there's a conflict in the zero-G situation. As I recall, in the gravity situation, any tilting movement—for example, the otolith organ response is such as would reinforce the semicircular canal movement, but this would mean then that in the space station, I mean in the zero-G situation, what you'd have is simply a reduction rather than a conflict.

Lansberg: If the brain stem is accustomed to receiving the combined information from semicircular canal and otoliths, this very lack of otolithic reinforcement can be just as disturbing as an abnormal signal. Fernandez and Alzate, in this country, have proved that, by cutting the utricular nerve in the cat. If this cat were put on his side, a paroxysmal nystagmus would occur, doubtless accompanied by a considerable vertigo; mind you, in an experimental set-up where the only abnormality was an absent otolithic—utricular—clue.

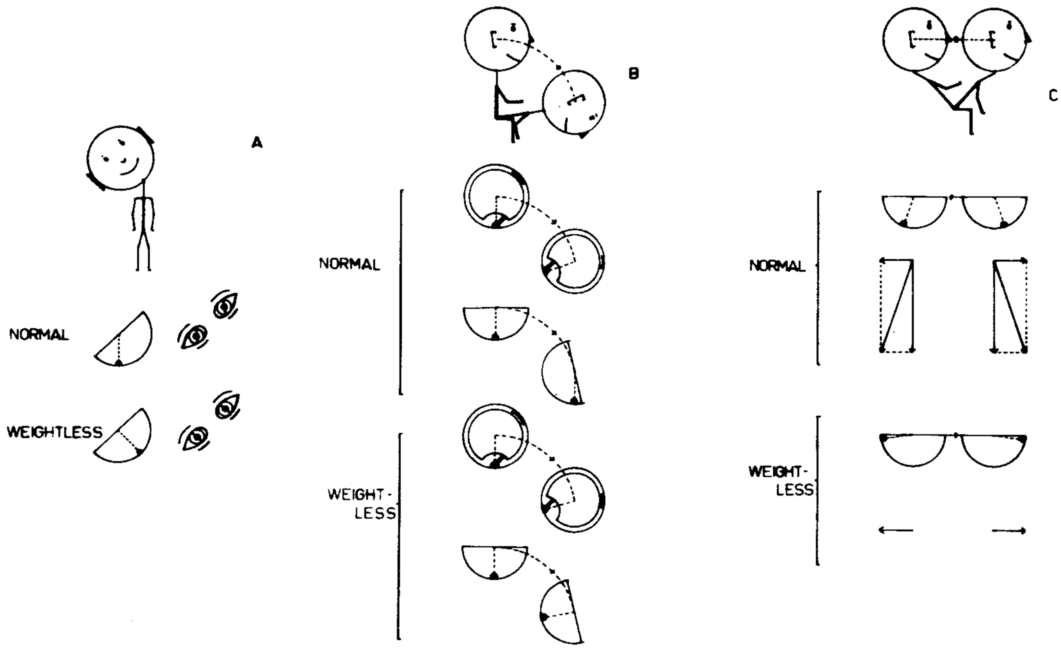

Figure 4. Sensory conflict during weightlessness.

a. Head inclined: Otolithic information remains nonetheless symmetric. Exaggerated representation of the counter-rolling eye movements.

b. Angular head movement: Upon initiation of the movement a cupular deflection is caused by the inertia of the endolymph. Termination of the movement drives the cupula back to its zero-position or slightly over. Drawing exaggerated. Otolithic signal inconsistent with cupular information.

c. Linear head movement: The otolithic information would suggest that the movement occurs in a vertical sense, which does not tally with other clues.

Hardy: I don't know those experiments you referred to—about how long after the operation was the experiment done?

Lansberg: Well, I think considerable time was lost. Much the same thing actually happens in Graybiel's slow-rotation room experiments. Head movements elicit vertigo and nystagmus until adaptation has occurred, but then, when the room is stopped after four days, the same head movements will again result in vertigo and nystagmus, in the reversed sense to be sure. It shows that a discrepancy between canal and otolithic impulses is liable to produce vertigo and nystagmus—a discrepancy which was either brought about by surgical interference (Fernandez, Alzate) or by functional interference (Graybiel, Guedry).

Dr. von Gierke: According to your explanation or hypothesis, you would expect the effect to be more severe in the beginning of the weightless state. Do you

have an explanation why Titov got along nicely for seven hours, and then suddenly the disturbances started?

Lansberg: Well, I think that for one thing motion sickness needs time for coming up but sometimes it's pretty quick, and it is quite possible that Titov felt somewhat strange during the first hours and kept everything quiet, and then, once getting accustomed to the situation, thought, "Well, I might move around a little bit." I don't know, the Russians were not quite liberal in telling me those things. Perhaps they don't even know yet; they said that after five hours it started and got worse, much worse until he had slept, and then it was over.

Dr. Hardy: Thank you very much, Dr. Lansberg.

Lansberg: There are two more pictures on a quite different thing. We meant to find a more certain way of predicting the incipient gray-out or black-out. We decided to do it in the following way: the man has a normal yellow center light as well as two red peripheral lights. He is instructed to fixate the center light until the peripheral lights are presented. At that moment he must direct his gaze to the red peripheral light, then to the left peripheral light, then back to the center light, and then press the button to put off the peripheral lights. The whole procedure will be seen from Figure 5. In Figure 6 the effect of G upon the ocular movement is demonstrated. At

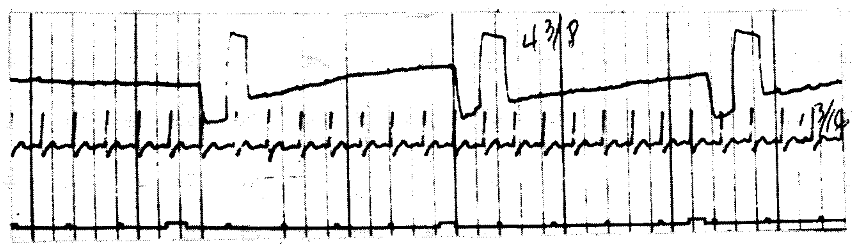

Figure 5. Recording of a standard ocular movement: The gaze is directed successively from the center to the right light, to the left light, and back to the center.

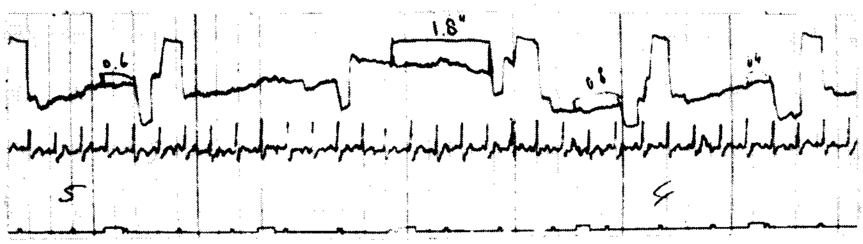

Figure 6. Same as Figure 5 but now at elevated G-values. Note the interrupted stroke as well as the increase in reaction time.

five G the gaze is no longer turned from left to right in one single stroke. One or more notches appear, indicating that the eye movement is now a stepwise one. As a tentative explanation, we theorized that, in order to direct the gaze from one peripheral light to the other in an uninterrupted

stroke, the image of second light should be present in the peripheral visual field right from the start.

Dr. Hardy: Well, we'll go ahead with the discussion now on both points of Dr. Lansberg's paper. Anybody have any further questions either about the ocular-gravic situation or about the objective method of evaluating G-tolerance and the positive-G position. It certainly seems as if this might be an improvement over the normally rather subjective course; this is subjective too, but I mean the evaluation depending on the subject to hold the fixed point. I think everybody agrees generally that the untrained subject will not do, and even the trained subject has some difficulty with this. There's no technical difficulty passing these electrodes as far as I know, and I think most American centrifuges are fitted out also with lots of recording channels.

Any comment on this? This has been a topic of great and sustained argument among many people for a long time—how one should evaluate G suits and what one should use as a criterion. In this last picture (Figure 6) I wasn't so sure that the particular recording that would have interested me you didn't comment on; that is, the one in which the chap apparently looked to the right and then looked back to the center and couldn't go any further; he just stopped, and that seems to me to be a possible endpoint. I'm not sure that it is actually a failure of vision when he looks to the right and then back and then has a hard time looking over to the left; I'm not sure this is a visual phenomenon or simply the fact that the muscles of the eye are not functioning as they do under one G.

Dr. Frank, would you?

Frank: The point is, of course, if you're going to turn his eyes, he mustn't turn his head. Do you fix the head?

Lansberg: Oh well, if he moves his head, then there wouldn't be a response like that. You always know that, and with closed-circuit television you see that he is not moving his head, but you would not even need that direct observation, because when he moves his head there is no response in the electric recording.

Hardy: This might be what he did in that middle recording.

Frank: Yes, very difficult to move your eyes without moving your head some. That's the normal response.

Hardy: This would take a little bit further work before it could be adopted as an international standard, but I know that at Johnsville this had been at least looked at as a possibility—not using actually electrical recording so much as using television monitoring. I think perhaps this is even a little better.

Lansberg: Apart from the pattern of the ocular movements, we were also interested in the effect of G first upon the electrical potential, and second upon the reaction time. We thought it might influence the amplitude of the electric potential at black-out level, but it does not. There is no change in electric potential that could actually have been predicted a little bit; Lewis and Duane have already performed experiments on the electro-retinogram with

no change occurring at black-out level. The reaction time is getting much longer.

Hardy: It may be quite a useful technique for further study.

Lansberg: It's just beginning.

Hardy: Thank you very much, Dr. Lansberg.

TABLE 1

	Period in sec.	Amplitude in M. A	Velocity in M/sec. A	Acceleration in $M/sec.^2$ $\omega^2 A$	Change of acceleration in $M/sec.^3$ $\omega^3 A$
SHIP = 1/2	13	20	10	5	2.5
SHIP = 1	6	2.5	2.5	2.5	2.5
TRAIN = 2π	1	0.01	0.06	0.39	2.5
SPRINGING = 4π	1/2	0.1	1.20	16	200

TABLE 2

	Cupula Deflection
Ship's rolling (heavy weather; double amplitude 15^o-period 18")	0.25^o
Space station (head rotation over 180^o) (rotating speed of space station 2/7 radian/sec.)	3.3^o
Barany test (10 rotations in 20")	18^o
Sinusoidal movement of the head (double amplitude 180^o) (period 1")	49^o

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FRENCH OBSERVATIONS AND RESEARCH CONCERNING IMPACT AND CRASH

by Lt-Col. Francois Violette
Centre d'Enseignement de Recherches de Medecine Aeronautique—Paris

Only few research experiments have been carried out in France, relative to impact, crash, and ditching problems. And presently no institute, civilian, military, or commercial, has any installation adaptable to carrying out similar research. Nevertheless, the French Air Force Medical Corps has been very much interested in those researches and installations. Since 1950, installations allowing linear deceleration studies have been planned and annually requested. But for financial reasons, they still have not been granted. Nevertheless, we hope to be able to build suitable installations in the near future at the Mont-de-Marsan Test Center. As for car constructors, nothing has been done, in spite of individual efforts of surgeons specialized in traumatology, or of adviser-physicians who are conscious of similar serious problems. Civilian physicians who had to answer questions dealing with occupant protection had recourse only to bibliographical sources and experimental research carried out in other countries, especially in the United States. In the field of car traumatology, as early as 1951, Petrignani pointed out the feasibility of protection, but little has been done, and it is only in a few cases that the feasibility of providing French cars with safety harnesses and belts is considered. Education of the people is just beginning. Only Air Force members who know about crashes—and not all do—tend to equip their seats with safety belts. A "Traffic Medicine" Society has just been created in France. As an active member of this Society, I regret saying that it is more directed to surgical therapy than to prophylaxis. The only attempt in prophylaxis has been maximum-speed limitation. Statistically the results have not been significant. Nevertheless, I think that with time we will be successful in establishing that prophylaxis is at least as necessary as therapy. I hope that in the next 10 years all French cars will be de-lethalized and supplied with safety belts and harnesses.

Such is the general situation in France at present; it is still far from brilliant, but is improving gradually. Already instrument panels are de-lethalized and external accessories liable to cause injuries have been forbidden. Now there is a favorable tendency toward safety belts. I hope that they will rapidly become compulsory and I am looking forward to government action to achieve this protection measure.

In the field of research applicable to aeronautics, the lack of specialized experimental means has been an important embarrassment. Nevertheless, this has not prevented us from studying a certain number of particular problems and from finding solutions. We shall consider successively: the problems, the study means, the results obtained, and the general trends. We bring only a modest contribution, but hope nevertheless that it will be somewhat useful.

Problems

Various problems relative to impact and crash have claimed Air Force Medical Corps attention, and attempts made to solve them with means at its disposal.

1. The problem of restraint of a wounded patient on a stretcher in case of crash of Medical aircraft has been examined. It is considered that a medical aircraft is covered by the Geneva Convention, and during war-time should run no risk. We thought it could occur that a medical aircraft could be attacked and damaged, and then crash. In order to avoid any untoward event, we decided to study the problem of sufficient restraint of wounded, in case of a crash, suitable for the highest number of wounded and non-ambulatory patients.

2. The problem of choice between ejection seats with solid axis guns and telescopic axis guns was a topic of discussion. This problem, which could have been solved by simple mechanical considerations, gave place in France to serious controversies. It also enabled us to study traumatology particularly with respect to ejection, and to differenciate ejection and impact injuries.

3. A very curious accident occurred during an emergency ditching of a helicopter. This accident, though exceptional, may occur again. Research has been carried out in order to protect flying personnel against such an accident.

4. Until recently in the Air Naval Command, conventional planes took off and landed with opened canopy, were of lower density, and in ditching floated several tens of seconds. Jet planes now take off and land with closed canopy and have such a density that they sink rapidly. The rescue of the pilot then necessitates ejecting through the canopy. This problem can be related to impact problems. Each aircraft type presents particular problems, and it was necessary to study emergency egress from French aircraft.

Study Means

The failure of specialized devices to simulate crash and impact conditions has not prevented different French scientists from attempting to study impact and crash problems.

Fabre utilized pathological methods; that is to say, observation and comparison of lesions caused by various accidents. This method permitted drawing generalized conclusions from numerous cases. In another case Mases and Jacquemin (1957), lacking information on decelerations observed in crashed planes, improvised experimental devices. After determining that abrupt braking of either cars or planes was insufficient for the purpose of this research, they organized collision of railway goods-trucks and even were successful with braking accelerometers. Finally, the only available device was a tip truck bumping strongly against an undamped buffer stop to obtain an acceleration peak.

Cabanon (1961) utilized British facilities for studying egress from the ditched Etendard IV plane.

Results

1. The study of restraint in crashes gave the following results, for patients or wounded carried with feet forward on longitudinally set stretchers. Two straps are sufficient: one on the base of the thorax, the other on the middle of the thighs. These straps, usually moderately tightened, must be sufficiently tightened in case of crash to prevent any slipping of the wounded on the stretcher. The restraint may be

comfortized: a) with stoppers placed on the arms of the stretchers, and b) by broadening the straps (three inches or 7.5 cm).

A quick strap release must be designed for the wounded. We recommend a release-pin system.

2. The statistical study showed us the superiority of the telescopic ejection seat, unfortunately with the occurrence of numerous vertebral accidents due to the solid-axis gun-ejection seat. It also showed that the breaking-point of the spine was about the seventh thoracic vertebra.

In the field of impact, the statistical study of ejection showed us the uselessness and danger of a helmet chin-strap in high-speed ejections.

3. An air naval jet-helicopter, having had an early flame-out while in the first "zone of death," ditched vertically but with moderate speed and fell flat on the water. The impact shock on the cabin floor was not felt by the standing crew members; nevertheless the pilot and co-pilot, although well strapped according to the rules, both received spinal fractures. The problem of the suitability of the strapping in such a case was thus presented, together with the anti-impact protection-device problem. We thought that in such accidents the impact-deceleration was transmitted without softening from the floor to the seat and to the strapped subject.

Our research is oriented towards shock-absorbing cushions. The event in question is uncommon but we think it possible in other countries. That is why we quote it.

General Trends and Conclusions

As long as we have no devices enabling us to study impact and deceleration, it will be difficult to establish a program. Nevertheless, we are planning to bring a contribution to the theoretical study of the impact effects, if possible. Our tendency in research is parallel to that of United States scientists.

Until recently, we knew only the laws of impact-response established in the 17th-century. The necessity of protecting the human being against shocks, but also of utilizing shocks to protect him, led to research initially empirical but now approaching theoretical explanations.

We express our gratitude and admiration to those who thus added to our knowledge and to the protection of flying personnel; and we are anxious to bring our contribution in this field.

AVIATION CRASH INJURY RESEARCH—REVIEW OF THE FLIGHT SAFETY FOUNDATION IMPACT WORK AND PLANS

Merwyn A. Kraft
Flight Safety Foundation

At this stage of these proceedings it seems appropriate to give recognition to the man who is probably most responsible for initiating the modern concept of human tolerance to crash forces and the subject of my comments, "Aviation Crash Injury Research." I refer to Hugh De Haven, who presently enjoys a greatly deserved retirement overlooking the Connecticut River at Hamburg Cove, Connecticut.

The concept of what was later to become known as aviation crash injury research was a result of a mid-air collision. As one of the pilots in the collision, De Haven received multiple fractures and internal injuries. The other pilot received only minor injuries. Both were regarded as lucky to have survived at all.

Upon leaving the hospital De Haven was given a desk job which included the handling of paper work related to all training-school accidents. Soon he began to notice that other injured pilots had the same highly localized pattern of injuries that he had experienced; also, that the severity of an accident did not necessarily dictate how severe injury was going to be. Men were being killed in relatively undamaged aircraft, while others were surviving in accidents where planes were virtually demolished. These observations led him to the belief that crash forces reaching a pilot were influenced or governed by the characteristics of the aircraft, such as the capacity of wings, landing gear, and other structural elements to absorb energy. Why wouldn't an extensive study of accidents bring to light the nature of these lifesaving characteristics which then could be built into future planes?

Even though no one would give very serious thought to this idea, De Haven continued to expound it. He also began to gather records on free falls where persons survived and, through contact with individual CAB investigators, accumulated considerable information from individual light-plane accidents. It was not, however, until the late 1930's that he was able to attract sufficient attention to obtain the support needed to inaugurate a broad study program. This came initially from Dr. Eugene F. Dubois, Chairman of the Committee on Aviation Medicine of the National Research Council, and head of the Department of Physiology at Cornell University Medical College. Dr. Dubois was impressed with the evidence accumulated by De Haven on the ability of the human body to withstand force, and brought this to the attention of the National Research Council. He also supported the idea of aircraft-accident investigation from a crash-injury point of view.

The assistance which De Haven received from Dubois did not stop with his calling the work to the attention of the National Research Council. Dubois provided De Haven with working space in the Department of Physiology at Cornell University Medical College even before De Haven was officially employed in April 1942 as research associate. Since no funds were yet available for De Haven's studies, the cost of the early

operations at Cornell were met on Dubois' recommendation by the Department of Physiology. Dubois continued to underwrite De Haven's work with funds from his own department until other sources of support could be found, and personally assisted in the early experiments performed at Cornell Medical College. The first funds specifically marked for support of the crash-injury-research project came from the Office of Scientific Research and Development.

De Haven used every opportunity possible to expound on his findings and theories. He presented many papers to the leading aviation societies such as IAS and SAE. He would buttonhole anyone who would listen to him, especially representatives of the airlines. As a result he received encouragement from the domestic and overseas airline operators in the form of letters of commendation to the Safety Bureau of the Civil Aeronautics Board.

It also is significant to note that the first official governmental acceptance of this crash-injury concept came while the present managing director of the Flight Safety Foundation, Jerome Lederer, was Director of the Safety Bureau of CAB. In July 1942, Mr. Lederer wrote De Haven enclosing a press clipping in which the CAB announced that its Safety Bureau had embarked on a research program designed to secure information from accidents in order to reduce the possibility of injury to occupants of aircraft. This marked the first attempt by an official organization to gather data for purposes of crash-injury investigation.

When financial support from the Office of Scientific Research and Development terminated in 1945, various other agencies took over. Since then funds have come from the Army, the Navy, the Air Force, the Civil Aeronautics Administration (predecessor to the FAA), AOPA, ALPA, several domestic and foreign airlines, and seat and aircraft manufacturers. Today the major funding comes from a contract with the U. S. Army Transportation Research Command and from a grant by the National Institutes of Health of the U. S. Public Health Service. Limited additional funds come from industry.

Initially, Aviation Crash Injury Research (AvCIR) was a unit of the Cornell University Medical Center. Later it was transferred to the Cornell-Guggenheim Aviation Safety Center, then to the Office of the Vice President for Research. On April 1, 1959, AvCIR, and its contract with the Office of Naval Research, was transferred from Cornell University to the Flight Safety Foundation. In September 1959 it was given formal status as a division of FSF. At approximately the same time the U. S. Army Transportation Research Command assumed the contractual obligations formerly held by ONR, and a program of reorganization and reorientation was started.

The research conducted by AvCIR is performed under the following categories.

Accident Investigation

This involves on-the-scene accident investigations of military and civilian, fixed-wing and rotary-wing accidents. The military accidents investigated are presently limited to Army aircraft which fall into both the light-plane and helicopter categories. Civilian accidents investigated include both transport and private aircraft in which AvCIR investigators serve as members of either the CAB or the FAA investigating team. This also involves evaluation of preliminary designs, mock-ups, and prototype and operational aircraft from a crash-worthiness or crash-injury point of view. Again both military and civilian aircraft are involved. The findings from these

activities form the basis for new improved designs and modification of existing designs from a crash-safety standpoint. The information compiled also becomes part of the mass data used by the organization for statistical studies.

Two instances, within the past year and a half, one civilian and one military, can be cited as excellent examples of the constructive interaction that comes from these field investigations. A popular civilian light plane was involved in two accidents of comparable configuration within a period of only ten days, both of which were investigated because they were in the Phoenix area, resulted in identical structural failures with crash-injury implications. A check of the previously reported accidents for this same make and model further confirmed the same inherent design deficiency. In due time, after this information was made available to the manufacturer, an effective design change was instituted. In the military case, involving a newly acquired aircraft still in production, major structural changes were incorporated into production models within a very short period after the results of accident investigation became available to the procurement office.

Statistical Analysis

These functions involve collection, coding, and analysis of data forwarded on special newly-designed AvCIR accident and medical report forms by the FAA, CAB, Army, State aviation groups, and State police forces. Statistical analyzes the mass data for the purpose of determining injury-causative factors. These analyses are published in statistical reports along with recommendations for corrective action regarding design features. Trend analysis measures trends of accident rates, injury severity rates, installation and utilization of safety equipment, and structural and design features related to injuries. Medical studies the frequency and pattern of injuries.

During the past year this phase of the program has been revitalized under the intensified guidance of Dr. Lee W. Gregg, Professor of Psychology at the Carnegie Institute of Technology, AvCIR's statistical and mathematical consultant, with the result that a new series of statistical reports has been published with papers accepted for publication in the Human Factors Journal, the Journal of Aerospace Medicine and the Journal of Environmental Health. Perhaps of even greater significance is the recognition given to AvCIR's statistical program by the U. S. Public Health Service. Just last week word was received that FSF had been given a two-year research grant formerly administered by the National Institutes of Health in the amount of $185,000, to follow up on current experimental statistical approaches and to initiate work on the construction of a predictive model in the form of a computer program. This is an ambitious program, and one which can make a major contribution to study in this field.

Training

Here attention is given to specialized training in the art and techniques of aircraft-accident investigation with emphasis on crash injury and survival. Currently five two-week courses are conducted each year for relatively small groups of 10 to 15 students. Those attending include both military and civilian design engineers, flight surgeons, aviators, safety officers, state police, state aviation personnel, representatives from FAA and CAB, and other similarly interested persons. To date a total of more than 200 persons have taken this course.

Human Factors

In this category the total fund of information relative to injuries and causative agents is correlated and interpreted in a clinical manner with emphasis on occupant restraint, occupant environment, protective equipment, and emergency evacuation. Recent studies have included "Impact Survival in Air Transport Accidents," "Mechanism of Aviation Crash Injuries," and "Limits of Seat Belt Protection During Crash Deceleration."

Experimental Research

This is the area of major significance for it is here that AvCIR is conducting full-scale crash tests of aircraft and the dynamic testing of components. This phase of the program was inaugurated in October 1960 with the drop of an H-25 Army helicopter from a height of 30 feet using a 75-foot crane moving at a speed of 28 miles per hour. Two anthropomorphic dummies were aboard, together with an experimental range-extender tank and a specially packaged oscillograph, altogether 34 channels of data were recorded, using both airborne and ground oscillograph equipment. (Following my remarks we will see a motion picture which will describe this crash test, and Dr. Turnbow in Dr. Lissner's panel tonight will give further details.)

Since that time four more helicopters have been crashed in a similar manner. An H13 and a HUP-2, (the Navy version of the H-25) were crashed in June of this year, and two similar helicopters in August. The basic purpose of these crash tests was to obtain acceleration and force data on the dynamic response of rotary-wing aircraft structure in three planes. The data will enable us to define the crash environment better for rotary-wing and VTOL aircraft. In addition, certain experimental modifications were introduced in each drop, so that each crash provides additional new information. For example, in the last HUP-2 drop, a new experimental troop seat designed on the basis of data obtained from the initial H-25 drop was incorporated to test a new principle. A satisfactory initial degree of success was obtained. Still another item checked was the energy-absorbing qualities of paper honeycomb under dynamic crash conditions. Again more details will be given by Dr. Turnbow.

Current and Future Programs

In addition to the expanded statistical studies, cited earlier, several new activities are underway. First, with funding support from the R & D Service of FAA, a study is being made covering the engineering concepts of a dynamic test device capable of producing realistic crash forces in three dimensions. A full report on this study will be available by mid-January, and preliminary discussion of it will be given by Dr. Turnbow this evening. Second, a special engineering study is being made of the crew and passenger restraint systems on all makes and models of Army aircraft. This includes a study of engineering considerations involved if seat belts and shoulder harness were to be anchored to basic structure. Since most Army aircraft have civilian counterparts, the results of this study should have broad application. As to the current crash-test program, six H-21 Vertol helicopters are on hand, three of which are scheduled to be crashed within the next eight months. The principal difference from earlier drops will be that all rotors will be in full operation so that there will be 100 per cent true simulation. The first drop will be strictly for post-crash fire research and, while that phase of our activity is not directly related to impact acceleration, it might be interesting to know that we are in the process of developing a fire-inerting and fire-suppression system for the H-21 helicopter. We will test out that

development while at the same time getting crash-injury information in these particular crashes. The third drop in this series will be a drone under remote control. We currently are developing a drone system for the H-21, which we hope will be available within six months. A variety of experimental installations will be incorporated in these particular crashes.

For the future, the AvCIR Division of the Flight Safety Foundation looks forward to continuing analysis of crash-injury and crash-survival factors, and to a long-range expanded program of dynamic testing of aircraft and components. Assurance of support also has been received that should permit the utilization of primates in early crash tests. This will be particularly valuable in assessing the effects from the relatively high vertical crash forces found in VTOL STOL aircraft accidents and expected in modern jet-transport cases. AvCIR and FSF welcome the opportunity to participate in this very valuable exchange of information. It is hoped that our findings will serve to support other constructive efforts in related areas.

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CAUSES OF IMPACT INJURY IN AUTOMOBILE ACCIDENTS

Robert A. Wolf
Director, Automotive Crash Injury Research
Cornell University

Introduction

During the early years of ACIR's growth we were primarily engaged in attempting to discover the various causes of injury to passengers in automobile accidents with not too much concern for the relative importance or the ultimate number of causes in the spectrum. Enough progress has now been made in identification of causes of injury that it is possible to review this spectrum and attempt to rank the discovered causes in their order of importance. A first paper on the topic of ranking causes of injury was presented at the Fifth Annual Stapp Conference on 14 September 1961; that paper was concerned mainly with the development of a methodology of selecting major causes and ranking them[(1)]. Today's discussion is a continuation of the ranking investigations, with special emphasis on variations in rank with different areas of crash impact. The automobile is subject to crash impact from all directions; it is this characteristic which makes it so difficult to provide a good "passenger-packaging" solution. There are other non-technical difficulties to "packaging" which plague the automotive safety engineer, such as styling, public acceptance, cost limits, etc. However, from the viewpoint of technical engineering difficulty, I suggest that the infinite direction of impact forces experienced in automobile accidents and their wide variations in magnitude pose one of the most baffling problems in arriving at practical solutions to good packaging.

For this reason it seems appropriate today to emphasize the effects of differing directions of crash impact and show some of its influence on injury causation. A ranking procedure is used as a convenient scheme for displaying these effects in a common framework which, it is hoped, will lend perspective to the overall pattern. Since this symposium is interested in the general nature of the work being performed by different organizations engaged in research in impact stress and injury, the ranking process also provides an overview of the type and scope of work being performed by ACIR in one of its major areas of inquiry.

Ranking causes of injury has several advantages:

1. It provides to the automobile designer an ordering of the prominent injury-producing features of the vehicle so that, together with other considerations, he can establish priorities for corrective-design action.

2. It helps to guide our own research program into areas of highest potential pay-off or overlooked regions of investigation.

3. It can provide, if time is introduced as a variable, a crude means of evaluating the effectiveness of safety-design changes as they are applied to production automobiles.

Although the ACIR project is concerned with three distinct areas of study—cause of injury, the nature of injury, and evaluation of safety devices applied to cars—this paper touches on only the first one, the causes of injury. These causes are discussed in Part I of the paper.

Part II of this presentation deals with some of the specific studies which ACIR plans to undertake during the next year and outlines some new areas of research related to its longer-range objectives.

I. Ranking Causes of Injury According to Directions of Impact

Definitions and Assumptions

A "cause" of injury, as used here, is a physical feature of the automobile which is the principal source of injury to a major body area of the passenger because of direct contact. Where the injury-producing contact occurs outside the car as the result of ejection, the ejection itself will be taken as the cause. In our data-collection process we have established 30 categories for classifying injury causes, but for purposes of tractability I will consider only 10 major items. These causes, listed alphabetically, are:

Backrest of Front Seat—Lower
Backrest of Front Seat—Upper
Door Structures
Ejection
Front Corner Post
Instrument Panel
Rear View Mirror
Steering Assembly
Top Structures
Windshields

The severity of injury sustained by passengers is an important variable, and the term must be defined before we can order the causes. The wide variation in degree of injury encountered in automobile accidents is so variable that a person can find a case to demonstrate almost anything if he has enough cases to choose from; thus, a statistical approach is desirable. Injury has many gradations ranging from a minor laceration to fatality, and for purposes of this study a subjective scale of four degrees of injury is established as follows:

Minor	Minor lacerations, bruises, fractures
Non-dangerous (to life)	Severe lacerations, fractures
Dangerous (to life)	Internal, brain injuries, etc.
Fatal	Death within 24 hours

Formulation of detailed rules to aid accident investigators in making consistent judgments of these degrees of injury and assigning them to causes is part of the regular indoctrination phase of ACIR field-data collectors. A check of these judgments is also made by ACIR Case Analysts as each case is reviewed in New York before encoding in the IBM card system.

The main variable to be discussed is direction of impact; six categories will be used—ALL, FRONT, SIDE, REAR, ROLLOVER and OTHER. Figure 1 indicates the frequency of impact occurrence from the Front, Side, Rear, and Rollover as it occurred in about 10,000 cars involved in injury-producing rural accidents during the period 1956-1959. Note that forward-quarter impacts account for about half the total. Side impacts are about 15 per cent with more right side than left side impacts, and rear impacts of sufficient severity to produce injury are only five per cent of the total. Note that rollovers are the second highest in occurrence, accounting for 23 per cent of the sample. The arrow at about ten o'clock does not indicate the direction of rollover impact; principal rollover generally results in a combination of side and top impact damage to the vehicle.

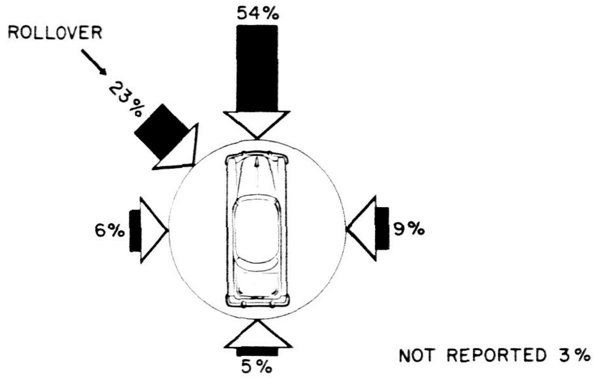

Figure 1. Direction of principal impact (% of cars—IP 1956-59 sample).

S[illegible] Illustrations

Before proceeding with the ranking analysis, I should like to illustrate a few cases of impact direction, degree of injury, and causes of injury from actual accident records.

Figures 2 and 3 show a 1959 Plymouth Plaza two-door sedan which was traveling at about 55 mph and struck a concrete abutment at a toll barrier. The driver said that

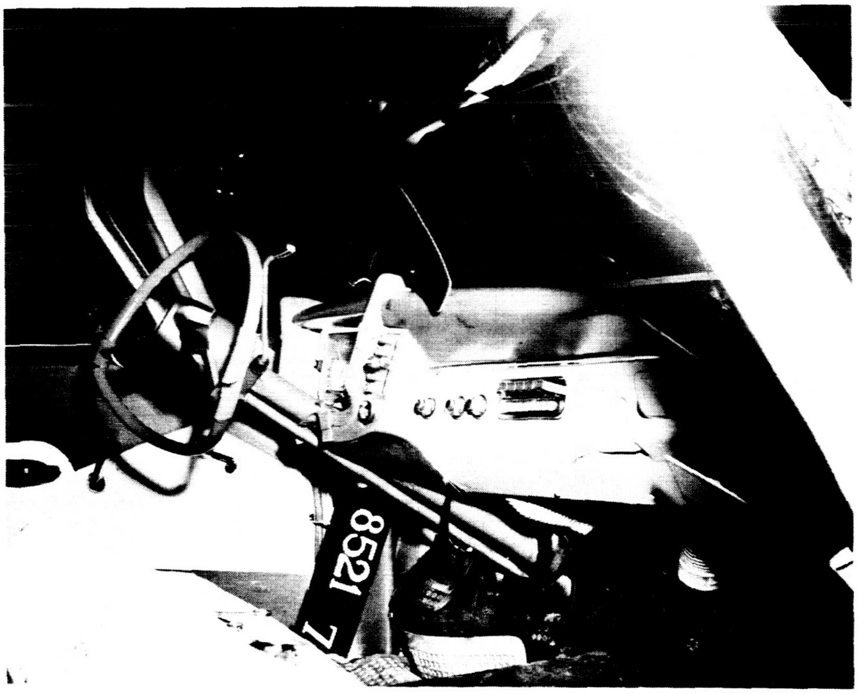

Figure 2

Figure 3

his brakes failed. Three persons were in the car: the driver, a man about 50 years of age; right front passenger, a woman about 50; left rear passenger, a woman in her late twenties. The impact was on the left front corner of the car. Injury was as follows:

Passenger	Cause of Injury	Degree of Injury
Driver (Chest struck wheel)	Steering Assembly	Non-dangerous (fractured ribs)
Right Front (Head struck windshield and instrument panel)	Windshield and Instrument Panel	Non-dangerous (cerebral concussion and broken jaw)
Left Rear (Hurled into right front compartment)	Backrest of Front Seat—Top	Non-dangerous (contusions to both legs)

In this case we have three passengers, four causes, and one degree of injury.

Figure 4 shows a 1957 Chevrolet four-door sedan which suddenly skidded, crossed a road divider, hit a bridge, spun, hit the bridge again, ejected the right front passenger, and came to rest. Two passengers were in the car: the driver, a woman about 50 using a seat belt, and a right front passenger. This passenger, a woman about 50 but not using her seat belt, was killed. Impact was from different directions and at different times. Injury was as follows:

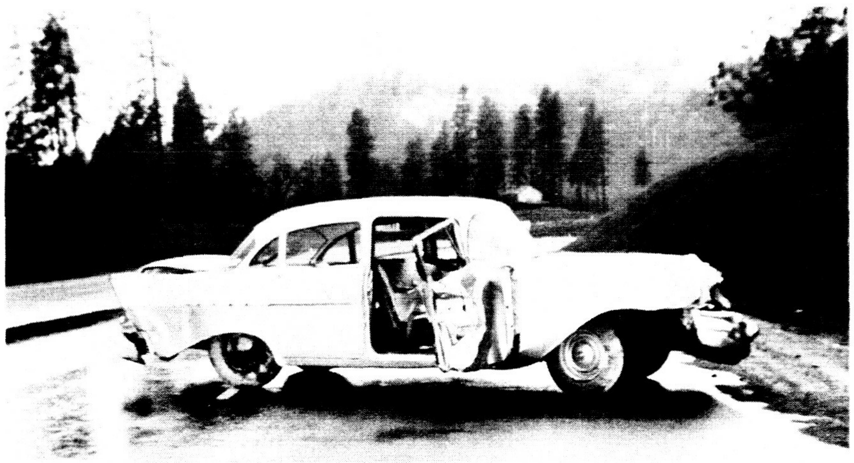

Figure 4

Passenger	Cause of Injury	Degree of Injury
Driver (Wearing seat belt)	Non-specific	Minor (bruises and lacerations)
Right Front (Seat belt not worn)	Ejection	Fatal (fractured skull, broken neck)

You have noted from the previous discussion and the illustrations that there is great variability in accident severity, direction of impact, force of impact, and degree of injury. These illustrations also show the widely different injury patterns occurring in the same vehicle.

Ranking the Causes

I am next going to discuss a group of injury-producing accidents involving cars manufactured during 1956 and later. There was a total of 26,131 passengers in these accidents, of which 19,356 were injured. The ranking which follows is concerned with the causes of injury to this sample of passengers. I should first like to indicate the priority of the ten injury-producing features without separation of impact directions; all directions of impact and rollover are lumped together to form a category labeled "All Impacts." Following this I shall discuss the changes of ranking of the causes in the separate categories of Front, Side, Rear, Rollover, and Other impacts.

Rank for All Impact Directions

Table 1 shows the data for the whole sample called "All Impacts," and indicates gross ranking of causes of injury according to per cent of total occupants receiving any of the four grades of injury. All directions of impact, including principal rollover, are lumped together in this table. However, the degrees of injury are separated so that one may observe the number of injuries in each category and also recognize the tendency for more frequent occurrence of minor injuries than dangerous to fatal grades.

TABLE 1

Number of Occupants Injured by Leading Causes of Injury
All Impacts
(IP 1956+ Cars)

Causes of Injury	Degree of Injury					
	Any Inj.	Fatal	Dang.	Non-Dang.	Minor	Deg. N.R.
Instrument Panel	4608	86	109	984	3182	247
Steering Assembly	4140	133	199	960	2461	387
Windshield	2943	84	92	599	1842	326
Door Structures	2418	62	75	544	1586	151
Ejection	2216	278	337	599	858	144
Backrest of Front Seat (Top)	940	9	40	187	597	107
Top Structures	782	48	49	120	448	117
Backrest of Front Seat (Lower)	726	1	18	134	533	40
Front Corner Post	496	51	34	122	240	49
Rear View Mirror	480	7	8	87	327	51

Minor injuries account for about 60 per cent of reported injuries while fatalities are about four per cent. The numbers in the first column labelled "Any Injury" are, for a given cause, the sums of the five components of injury in that row. For example, adding the first row from right (247) to left (86) gives 4,608 injured by instrument panel. There is a great deal of interesting information in this table which may be subject to further analysis. For example, if one were to rank the causes of injury according to number of fatalities, rather than any injury, then ejection would be the highest ranking cause, followed by steering assembly. The gross form of ranking shown on this table is useful in that it yields clues which may point out areas for further study and eventual corrective action in automobile design. An indication of such a clue is given by the fact that in an earlier and smaller sample of data, collected in the pre-1956 period, ejection ranked second, while in this sample of post-1956 cars it has dropped to fifth place. This change is probably due to the effectiveness of the new safety door locks which were introduced by the automobile manufacturers in the post-1956 models. A separate study probing into this problem in more detail has recently been completed by ACIR[2]. Another point of interest is that the first five causes of injury appear to fall into a grouping of strong influence compared to the weaker effect of the last five.

Discussion of the detailed effects of direction and type of impact as they influence injury is of more specific interest to the designer. The following tables indicate causes for each principal direction of impact, and these causes are again ranked on the basis of per cent of total passengers injured by the indicated cause.

Comparison—Front and Side Impacts

Table 2 shows ranked causes of injury for two directions of principal impact, Front and Side. These may be compared to the "All" category which is repeated from Table 1 for purposes of convenience. The number of passengers for front-impact accidents is 12,252, while those for side-impact accidents are 3,345. The percentages shown are determined by the number of reported injuries attributable to each cause

TABLE 2

Per Cent of Occupants Injured by Indicated Cause
(IP 1956+ Cars)

Area of Principal Impact					
All Areas	%	Front	%	Side	%
Instrument Panel	17.6	Instrument Panel	26.1	Door Structures	19.7
Steering Assembly	15.8	Steering Assembly	23.6	Ejection	13.3
Windshield	11.3	Windshield	15.1	Instrument Panel	12.0
Door Structures	9.3	Door Structures	6.0	Steering Assembly	10.2
Ejection	8.5	Backrest of Front Seat (Top)	4.1	Windshield	7.8
Backrest of Front Seat (Top)	3.6	Backrest of Front Seat (Lower)	3.9	Front Corner Post	3.0
Top Structures	3.0	Ejection	2.9	Backrest of Front Seat (Top)	2.7
Backrest of Front Seat (Lower)	2.8	Rear-View Mirror	2.4	Top Structures	1.8
Front Corner Post	1.9	Front Corner Post	1.7	Backrest of Front Seat (Lower)	1.7
Rear-View Mirror	1.8	Top Structures	1.1	Rear-View Mirror	1.3

after division by the number of occupants within the specific impact group. Door structures are blocked out for particular attention because, although this cause is generally in third or fourth place for most directions of impact, in the case of side-area impacts it rises to first position of importance. This unusual shift leads to the question—Why? This puts the spotlight on an important subject for further study. Seated position of the passengers undoubtedly has something to do with it, as passengers sitting next to a door which is the impact point are usually injured more severely than others; this may be due to the invasion by the impacting car pushing the damaged door structure into the passenger compartment; it may also be due to the passenger impacting an undamaged door. Another point of significance emerges here in that the occurrence of ejection appears as a strong cause in side impacts (2nd place) even though it was rather low in front impacts (7th place). It is important that automobile designers note this low occurrence of ejection (door opernings) in frontal impacts, as many of the crash-test experiments conducted by the industry are head-on into barriers or other cars. The absence of door openings in frontal test impacts does not necessarily prove that door locks are adequate. This result would appear to indicate that side impacts would provide a better criterion for testing door locks. It must also be

noted that some side impacts, especially in the neighborhood of the door, cause so much buckling and deformation of the frame back-up structure or the door itself that the best locks cannot hold.

Comparison—Rear, Rollover, Other

Table 3, which compares rear, rollover and other impacts, is similar in construction to Table 2. The number of passengers in each group is as follows: rear 1,876, principal rollover 5,086, other 3,572. Instrument panel is blocked out for

TABLE 3

Per Cent of Occupants Injured by Indicated Cause
(IP 1956+ Cars)

Area of Principal Impact					
Rear	%	Principal Rollover	%	Others*	
Instrument Panel	6.9	Ejection	21.4	Instrument Panel	14.1
Backrest of Front Seat (Top)	6.3	Door Structures	11.2	Steering Assembly	12.5
Door Structures	5.0	Top Structures	9.0	Windshield	10.5
Steering Assembly	3.8	Windshield	8.2	Door Structures	10.1
Backrest of Front Seat (Lower)	2.8	Steering Assembly	7.6	Ejection	8.1
Ejection	2.3	Instrument Panel	7.4	Backrest of Front Seat (Top)	3.5
Windshield	1.8	Backrest of Front Seat (Top)	2.0	Top Structures	2.7
Top Structures	1.8	Front Corner Post	1.9	Backrest of Front Seat (Lower)	2.5
Front Corner Post	.4	Rear-View Mirror	1.2	Front Corner Post	2.4
Rear-View Mirror	.4	Backrest of Front Seat (Lower)	.9	Rear-View Mirror	2.0

special notice because, although it is a consistently high-ranking source of injury for all other categories of impact direction, it drops to a middle position in the case of principal rollover. In the violent tumbling of passengers in rollover, the instrument panel is only one of several objects which may be struck. It is also difficult for the accident investigator to assign a major injury to a particular cause in the rollover case—sometimes a car rolls two or three times.

Several factors about rear-impact accidents should be noted before comparing them with other types of impacts; the first is that a large proportion of the injury is in the minor category. Fatality is quite low—constituting about two-tenths of one per cent of occupants in the rear-impact group, compared with about three per cent of occupants for all areas of impact. Incidentally, the fatalities in this sample of rear impacts all occurred in the ejection category. Instrument panel in rear impacts produces a surprisingly large portion of injuries even though they are of a minor degree; more than 25 per cent of the minor injuries within the rear-impact group were produced by this single source. It is highly probable that most of these injuries occur in the passenger-rebound phase of motion or in a secondary frontal type of impact; more detailed study could clarify this point. The top backrest of the front seat also emerges as a strong cause of injury in the rear impact accident.

The most significant observation in the principal-rollover group is that ejection has moved up to first place; of the fatalities occurring in this group, about two-thirds were caused by ejection. Top structures have understandably moved up from their relatively low rank in the other categories of impact.

Since the impact category marked "Other" represents a mixture of directions and areas of impact, and includes secondary rollover which by definition is caused by a primary impact from an outside source, detail comparison of causes of injury in this group are not made. The rankings in this category are identical to the "All" group shown in Table 1.

<u>Putting Causes into Perspective</u>

Figure 5 brings the ranking of the five impact groups together so that one may trace the shifts of the causes of injury within the entire group. Each block represents one of the categories of impact previously discussed. The horizontal dimension of each block is proportional to the percentage of total passengers in that category of impact. These percentages are shown at the bottom. The vertical scale shows the number of reported injuries attributable to each cause after division by the number of occupants within the specific impact group. Thus, the area under the per cent line for a given cause would show the contribution of each impact category to the overall number of injuries attributable to that cause. The top-ranking cause is not at the same vertical level for all impact categories as each class has a different starting level for the top ranking cause; these are indicated by the abbreviated tick on the left border of each block. For example, IP (Instrument Panel) at 26.1 per cent for Front—DR (Door Structure) at 19.7 per cent for Side.

Only two causes are shown on this diagram for the sake of simplicity—steering assembly and ejection. Tracing the pattern of steering assembly summarizes its ranking as 2, 4, 4, 5, 2 for impact area of front, side, rear, rollover, and other. Thus, the steering column and wheel persist for all areas of impact as one of the five leading causes of injury. Since steering assembly also ranks second in the front-impact group which accounts for about half of the impacts, it should have high priority in redesign for safer automobiles. ACIR plans to perform a detailed study of steering-wheel configuration to determine the effect of recent variations in design, such as recessed hubs, two spokes, three spokes, etc. There is some evidence that steering wheels are in fact protective devices up to some level of impact severity and it would be enlightening to be able to determine at what point the wheel changes from a protector to a threat. An often-overlooked factor about design configuration must be pointed out here: Supposing one could, by redesign, entirely eliminate the steering

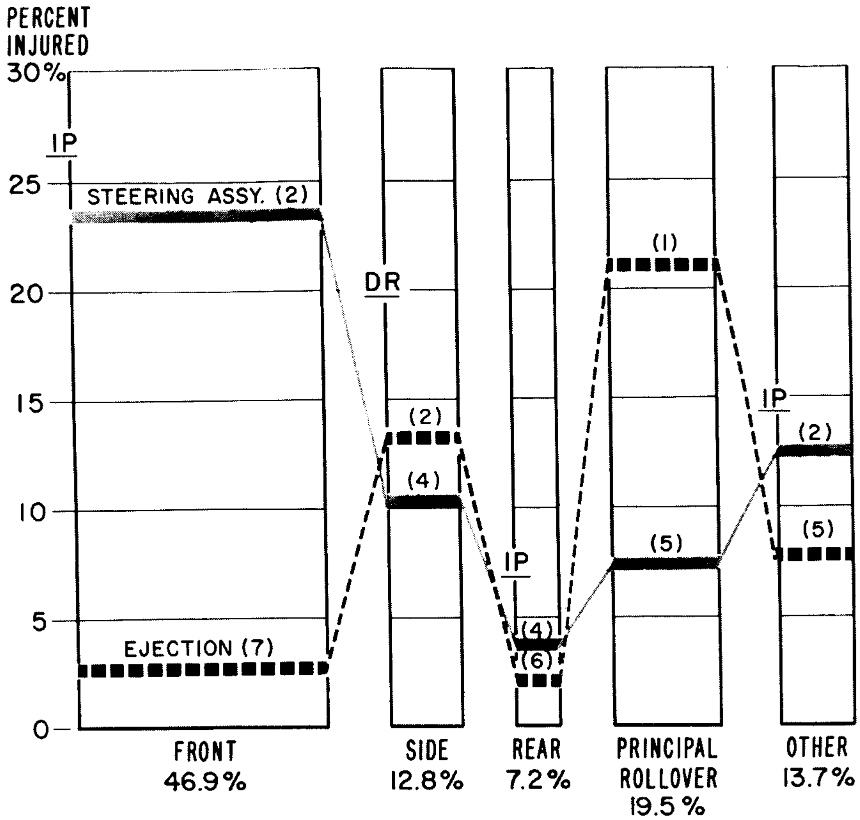

Figure 5. Risk by area of impact (IP 1956 + Cars).

assembly, would it eliminate all injury now attributed to it? If the wheel and column were removed, what kind of structure would face the driver? If it is a structure something like an instrument panel, we may expect injury of about the type and degree characteristic of the right front passenger position of our present cars, by considering the instrument panel as an injury producer. This highlights an important aspect of redesign in relation to injury causation. The designer must not merely replace one injury-producing feature with another equally as dangerous—the solution must actually be more fundamental, and reflect a real understanding of the possible injury-causing characteristics of the alternative structure. Carrying the process one step further—if the instrument panel is recessed or carried forward—again some other structural feature such as a windshield frame will still be present. What lethal implication will it have? This hypothetical procedure of removing the steering column only to be confronted by an instrument panel, then removing the instrument panel only to be confronted by a windshield and cowl frame, illustrates that simple elimination of any one cause of injury will not, of itself, necessarily eliminate all of the injury attributed to it. One solution to the problem is to prevent passengers from impacting these parts of the automobile. This is the role of the seat belt and safety harness and, in combination with it, recessing the instrument panel may provide clearance for the passenger when "jackknifing" or stretching the harness.

Ejection is also singled out as an important cause of injury. Although it is receding in importance in front, rear, and other impacts due to redesign of door locks in 1956 and subsequently, it is still a strong cause of injury, especially in side impacts and rollovers. This specifies the crash conditions to which new designs of door locks should be pointed. It may be assumed that control of ejection in this sample

results mostly from door locks, as safety belts were not used to a degree sufficient to have much influence on ejection control. The other causes of impact injury may be traced in similar fashion, but there is not sufficient time to discuss each cause in this paper.

Summary of Part I

The general process of data collection and analysis of automobile accidents as practiced by the Automotive Crash Injury Research project of Cornell University has been outlined. Although the project is generally involved in three distinct areas of study—causes of injury, the nature of injury, evaluation of safety devices applied to cars—I have touched only upon causes of injury. These causes, which are physical features of American automobiles built since 1956, have been discussed in the framework of a ranking procedure which provides a measure of the order of importance of ten major injury-causing features that have been identified by ACIR. The causes are discussed in terms of their variation with six categories of impact: All directions lumped; then front, side, rear, principal rollover, and other. Five or six of the ten features of automobile interiors emerge as consistent major causes of injury. They are instrument panels, windshields, steering assemblies, door structures, and ejection. Ejection is slowly dropping toward the bottom of the list as time passes. Some other causes appear dominant only under certain directions of impact; for example, top structures emerge as a strong cause only in rollover. The ranking procedure used here is not a substitute for detailed and controlled analysis, as it is essentially a tabulation process, but it has several important uses:

1. It provides an overview of the type of research which ACIR performs.

2. It provides some indication of an order of priority for the automobile designer in his efforts toward building safer cars.

3. It provides clues for guidance of further study and research.

The reader's attention is called to the fact that these results are based on accident-injury experience in the pre-seat-belt era in the United States, and that quite different results will probably be obtained as the use of seat belts becomes more prevalent.

II. Future Plans of ACIR

Discussion

In discussing plans for the future, I shall speak of ACIR as an organization in transition. It has been in existence for about eight years and during this formative period has intensively pursued a single type of research—injury causation—by means of a single basic process. It is now appropriate to review the past and consider the addition of several lines of complementary research designed to enhance ACIR's capability to perform useful service in the interest of improved highway safety. Such additional branches may be (a) in the areas of improved methodology directed toward the traditional area of Cornell's crash-injury research, figuratively "providing a higher-powered microscope" and (b) entry into the field of accident causation. I will discuss both of these branches very briefly and then outline some specific research tasks which may serve to illustrate our general aims and hopes for the future.

The early development of ACIR was motivated by an intense interest in reaching an understanding of the causation of injury to passengers once the accident process had started. It was believed that, if specific causes of injury could be identified, then corrective measures could be introduced by redesign of automobiles. The statistical-analytical process of probing into automobile accidents on a massive scale was chosen initially as a promising means of gaining understanding and a nationwide data-collection system was developed to provide the basic input data for study. As its new director, and taking a retrospective view, I believe that this early vision was sound, the approach was correct, and that the product of ACIR's research has been of benefit to society by adding understanding of injury causation and stimulating improvement in the safety of American automobiles. A great deal of credit is due to the vision of the originators of ACIR and the encouragement by its sponsors.

This research may also have influenced the design of safer cars in other parts of the world as we find considerable interest of international origin in our work. Tangible results of foreign interest are more difficult to evaluate than those actions taken in the United States but safety door locks are beginning to appear in foreign cars and instrument panel padding is becoming more prevalent. It appears to me that the future influence of this research should begin to have stronger international implications. Many other industrial nations are beginning to experience the sociological change and the traffic and safety problems which are created by an automobile-oriented transport economy such as we have in the United States. Many of us hope that we are learning to live with the automobile as well as by means of it—perhaps we can teach other nations some of the lessons we have learned.

It is my hope to see much more knowledge of the injury-producing process emerge from ACIR and applied to improving automobile safety. We hope to provide the designer with basic information not only on what features cause injury but also where are the best payoffs to reduce injury and what priority he should consider in applying his efforts. The present preoccupation of ACIR with ranking causes of injury according to their injury-producing importance is pointed in this direction. Whether one is engaged in basic or applied research usually boils down to a problem of semantics. I like to view our work as basic, that is, aimed toward gaining knowledge of the fundamentals of injury causation. However, we also have a keen interest in seeing this knowledge applied in actual practice. But we have no direct leverage to force application of our findings; this must be done by factual persuasion. Our work is mainly concerned with broad effects and trends. These frequently transcend detail variation in automobile configurations which occur from year to year and from manufacturer to manufacturer. We are fully aware, however, that details of design are extremely important, in fact highly critical in their injury potential and that year to year monitoring of safety design detail must be maintained by the automobile manufacturers. It is important, therefore, to realize that ACIR is interested in specific product evaluation only insofar as it may lead to general improvement in design; we are not engaged in rating name brands or evaluating specific models of cars.

In helping to bridge the gap between gaining basic knowledge and application of it to product design, I believe that ACIR can make further contribution by converting some of its findings into terms more directly understandable to the engineer. The mechanism would be formulation of performance requirements—documents combining ACIR research findings on causes of injury with engineering analysis and experiment pointed toward principles of solution.

I mentioned another form of activity which appears desirable to add to ACIR's present interest; this would be directed toward accident prevention.

Our approach to accident prevention would at first be sought through the medium of development of theories of accident causation. Very little research has gone into development of basic theory of accident causation, and there appears to be a great need to supplement the traditional empirical approach with integrated, quantitative theory. A first step in this direction would be an attempt to formulate theories of certain classes of accidents by means of mathematical models to simulate man-vehicle-highway system situations. Such models would be formulated to represent chosen highway intersections and would be programmed for analysis by means of digital computers. The influence of the many variables in the systems could then be explored in a systematic fashion. Considerable success has been achieved in mathematical modeling and fast-time simulation of military systems, and the experience of the Cornell Aeronautical Laboratory would be drawn upon in this effort. The modeling process would utilize and attempt to integrate much of the background developed by other research in the behavioral sciences, relating to driver tasks and driver behavior, in the engineering sciences relating to vehicle design and handling qualities, and in highway engineering and traffic control.

Specific Research Tasks

The specific problems listed below illustrate the overall program of ACIR; these are divided into three groups—studies in process, near-term planned studies, far-term aims.

Studies in Process

Speed effects in injury-producing accidents are being reexamined.

Seated position and seated patterns as related to injury are being updated as more information becomes available.

Effectiveness of instrument-panel padding is of high interest; comparisons in injury patterns are being made with unpadded and padded panels.

Occupant physical characteristics are being studied in relation to injury; main factors examined are weight, age, height, sex.

Top damage and injury patterns are being studied in rollover accidents.

Differences are being sought between injury-producing (IP) and property-damage (PD) accidents.

Ranking of causes of injury is a continuing process to show the relative importance of the various causes.

A seat-belt-usage survey is being conducted to show trends of public acceptance.

Near-Term Planned Studies

It is planned to undertake many of the following studies during the next year.

Safety-Belt Studies

Injury patterns of non-ejectees, with control of buffeting by means of seat belts.

Injury caused by safety belts—type and frequency.

Belt failures, incidence and type.

Comparison of the right front seat position in forward-impact accidents—with and without belts.

Effectiveness of safety-steering-wheel assemblies.

Safety glass—injury patterns and effectiveness.

Penetration by striking objects and resulting injury in side impacts.

Pilot study in truck involvements and injury patterns.

Pilot study in the potential use of conventional state-collected data.

Comparison of injury patterns in large American cars, American compact cars, and foreign cars.

A medically oriented study which may aid either research or emergency treatment.

Far-Term Studies

Some of these may be started within the next calendar year, if funding becomes available.

Long-range investigation of truck involvement and injury patterns similar to the passenger-car accident-research process of the past; this will require a new data-collecting system.

Continuing evaluation of safety features applied to new automobile design.

Changing injury patterns with changing design trends in passenger automobiles.

Comparison of different types of safety belts: lap, diagonal, shoulder harness, etc.

Engineering analysis of the mechanics of crashes to formulate theory and predictive methods for estimating impact forces on vehicle components and passengers.

Development of theory of accident causation by means of mathematical modeling and simulation of man-vehicle-highway system situations.

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HUMAN AND ANIMAL IMPACT STUDIES IN U. S. UNIVERSITIES

Herbert R. Lissner
Wayne State University, Detroit, Michigan

Current research in U. S. universities which is related to the effect of impact on human subjects falls naturally into three separate categories. Different aspects of the problem of human impact are found in each category, and investigation in all three areas must be vigorously pursued in order to obtain a complete and comprehensive solution of the problem. In the first classification we find the research which deals with programmed automobile crashes, either actual or simulated. In these carefully planned impacts, all of the details regarding acceleration, velocity, and trajectory of the vehicle before and after impact are accurately determined. In addition, anthropomorphic dummies are used to determine the possible trajectories of passengers subjected to the specific conditions of the vehicle impact under investigation. In these tests, in which all the details with regard to the impact are known, no information regarding the extent and severity of injury to the human are obtained, since presently available anthropomorphic dummies simulate the human only in regard to size and average-weight distribution. They therefore cannot provide information regarding the type and degree of injury produced. Extensive research in this category is being conducted by Derward Severy at the Institute of Transportation and Traffic Engineering, University of California, Los Angeles, and by James Ryan at the University of Minnesota, and much valuable information is being obtained from their work.

In the second category we have the investigations of actual on-the-road automobile crashes after they have occurred. In this classification we have extensive compilation of injuries produced in the accidents, and these are classified with regard to type and severity and related to the structural member or part of the car responsible for them. These studies furnish a complete picture of the extent and severity of injuries produced in various kinds of highway accidents and provide considerable information with regard to the specific structural part which caused the injury. This information is also extremely valuable, since it indicates the areas of the car which should be redesigned to provide greater energy absorption when struck by the body. Unfortunately, however, in this type of an investigation all data regarding accelerations, velocities, and energies responsible for the production of the injuries are lacking. Research in this category is being conducted by Robert Wolf and the Crash Injury Research Group at Cornell, which is collecting data from various parts of the country with the aid and cooperation of state and municipal police forces; and by Al Mosely and his organization at Harvard, which is making a very comprehensive investigation of every automobile accident resulting in a fatality in the Boston area.

In the third category we have the investigations being conducted primarily in the laboratory, which attempt to relate the extent and severity of the injury produced to the parameters of the impact responsible for the injury. In order to determine the extent of this kind of research being conducted in U. S. universities, a survey was made of 158 engineering schools and 83 medical colleges. Replies to a questionnaire were received from 84 engineering and 42 medical school deans, and I would like to

take this occasion to thank them for their cooperation. In many instances they obtained information for the survey from departments outside their colleges, which helped to correct a deficiency; that is, the questionnaire had not been addressed to departments outside of engineering and medicine, such as biology and veterinary medicine.

In the past, many tests have been conducted, primarily in medical schools, to examine the effect of impacts on animals and humans. In most of these tests, which were conducted before 1945, adequate instrumentation was not available to describe properly the impact that produced the injury. It has been only in the last 10 or 15 years that instrumentation has been available to describe the parameters of the blow adequately, and relate them to the kind of injury produced. Engineering-school faculties which would have been most able to measure and evaluate the extent of the impact generally have not engaged in research of this type. The current programs leading to advanced degrees in biomedical engineering at a number of schools in the country should be mentioned. The programs at these schools deal for the most part with electronics and its application to medical problems, primarily in the design of instrumentation. Programs are available at both the M.S. and Ph.D. levels. Among the schools with programs of this kind are the University of Pennsylvania, the University of Rochester, Johns Hopkins University, Iowa State University, and the University of Nebraska. None of these institutions is at present conducting any research involving impact to animals or humans.

The results of the survey of engineering colleges indicated the following projects in addition to those already cited. At Tuskegee Institute, an investigation is underway to study the electro-physiological effects in animals due to impact and vibration. At Northwestern University, Professor John E. Jacobs is conducting tests with the Northwestern University football team to determine the accelerations produced on the team members in action. Telemetering equipment is contained within the player's helmet, and information regarding the accelerations experienced and electro-encephalographic records are telemetered to recording equipment. Tests being conducted by the Mechanics Department of the Engineering College of Wayne State University, in cooperation with the College of Medicine in the Biomechanics Research Center, will be discussed in detail a little later.

In medical schools, current investigations involving impact are being carried on at the University of Texas, where Professor F. H. Rudenberg is conducting an experimental cerebral-concussion investigation on the rat. At the University of Michigan, Professor Donald F. Huelke is conducting investigations on the deformations of the mandible due to impact, using stresscoat to indicate the extent of the deformations produced. Dr. Reinhard Friede at the same institution is engaged in a project on cerebral contusions involving experimental work on cats. At Duke University, Drs. Norman Goodman and I. T. Diamond are conducting tests of impact and vibration on cats in the psychology laboratory.

Considerable information was obtained regarding the conduct of vibration studies at medical schools. At the Ohio State University, Dr. Walter F. Ashe is determining physiological and pathological effects of vibration in man and animals in a study supported by the National Institutes of Health. At Northwestern University, studies are being undertaken on the mechanical vibration of the brain with reference to the possible relationships to electro-encephalograms obtained during this vibration. Sonic vibrations are being used in research at the Carnegie Institute of Technology, Department of Psychology, at Villanova University, at the Radiation Biology Laboratory of Texas A and M, and at the Institute of Agricultural Medicine of the State University of Iowa.

Ultrasonic research is being conducted at the Yale University School of Medicine, the College of Medicine at the University of Utah, and at the State University of Iowa Hospital in cooperation with the Biophysics Research Laboratory of the University of Illinois. Professor R. E. Nichols of the Department of Veterinary Science, University of Wisconsin, is conducting research involving vibration of cattle to determine the effect of this vibration on shipping fever.

In the Mechanical Engineering Department of the University of Maryland, Walter K. Harrison, Jr. is conducting an investigation on the ballisto-cardiographic dynamics of the human body continuum. At the Wenner-Gren Aeronautical Research Laboratory of the University of Kentucky, Professor K. O. Lange is conducting research on the response of the human body to vibrations below 25 cycles per second. Research involving sonic vibrations is being conducted in the Department of Mechanical Engineering at the Massachusetts Institute of Technology and in the Department of Psychology at the Illinois Institute of Technology. Ultrasonic investigations are being performed in the schools of engineering at Vanderbilt University and at Manhattan College in the Department of Biology.

The response to the inquiry regarding the availability of equipment and personnel for the conduct of tests involving impact to animals and humans was very gratifying. Twenty-one engineering and nine medical schools indicated the availability of personnel and equipment and a desire to participate in tests of this nature. While the survey indicated a lack of specialized equipment available for the conduct of tests of this nature, it did show that personnel were available and interested in conducting such tests. It has been our experience that by far the most important aspect of conduct of research in this area is the desire of personnel to engage in such cooperative investigations, and for this reason it is felt that the response obtained was most encouraging. A list of the institutions indicating an interest in performing impact and vibration tests is presented at the end of this paper.

I would like to describe some of the special equipment that we have available at the Wayne State University Biomechanics Research Center for the conduct of controlled-acceleration tests on humans and cadavers. In an unused elevator shaft in the eight-story Medical Science Building, we have constructed a vertical accelerator. This equipment is powered with compressed air stored in a 35 cubic-foot capacity tank in the basement. The tank is connected to a cylinder five inches in diameter and eight feet long. A piston in the cylinder drives a carriage which is mounted on two vertical steel rails that run to the top of the building. A pressure of 1,000 pounds per square inch may be applied to the piston in order to accelerate the carriage, which holds an air-force-ejection seat shell in which a cadaver or animal may be mounted. A valve between the tank and the cylinder permits the rate of acceleration of the carriage to be varied between 50 and 2,000 g's per second. Since the stroke of the piston is only eight feet, the maximum acceleration that can be achieved at the lower rates of acceleration is fairly small. At the high rates of acceleration an essentially square wave pulse can be applied to the carriage, the magnitude of the acceleration being 50 g and its total time duration 1/10 of a second. Friction brakes on the carriage grip the rails and bring the carriage to rest before it reaches the top of the building. Current tests with this equipment, supported by the National Institutes of Health, are being run to determine the effect of the rate of acceleration on the level of the strain produced in the vertebra of the cadaver mounted on the carriage. Strain gauges are cemented to various vertebra along the spine, the leads are brought out to recording equipment on the fourth floor. The leads are securely fastened to the cadaver, and despite the fact that they

whip about violently at each test, they have not broken in any experiment to date. Figure 1 is a diagrammatic sketch of the accelerator.

The rails of the accelerator are also useful for guiding cadavers being dropped onto various test surfaces to duplicate the actual blows experienced in certain kinds of impacts. A current study involves dropping the cadaver to permit the head to strike several kinds of safety glass at velocities up to 35 miles per hour. Instead of applying a controlled-acceleration pattern to the head, the accelerations produced in this type of test are exact duplicates of those which exist when the human being impacts the structure at the same velocity. The interpretation of the results of these tests has been possible because of the theory which was developed for the mechanism of concussion. This theory states that the concussive effect is proportional to the shear stress developed in the brain-stem region of the brain. Since the shear stress is dependent on the pressure developed in the cranial cavity and its time duration, a measurement of these factors permits us to predict the degree of concussive damage that would result to the individual striking the structure. It is possible not only to indicate that this sort of impact would be lethal but also to determine the severity of the injury produced by a particular impact. We believe that this type of theory must be evolved for impacts in other regions of the body to determine the degree of damage produced by any particular blow.

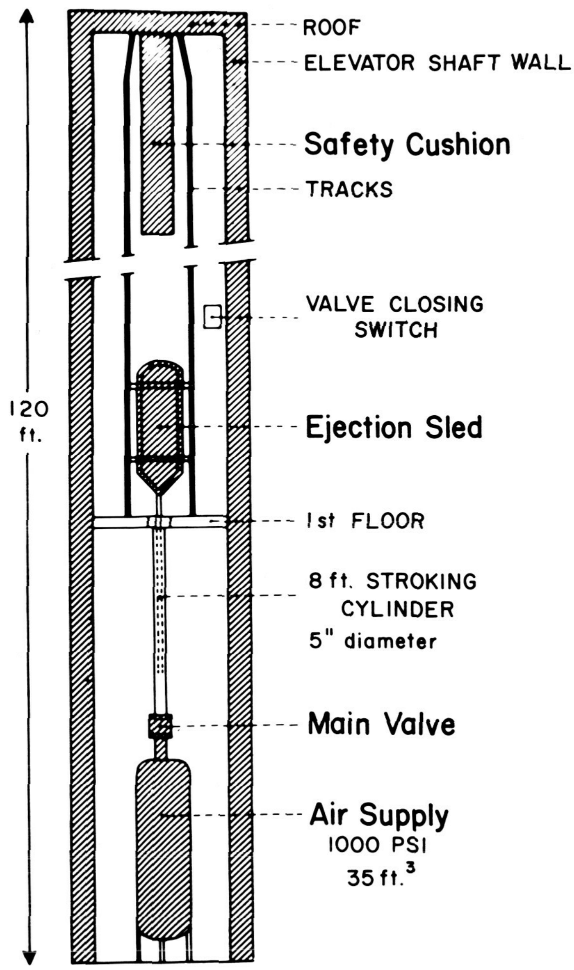


Figure 1. Wayne State University Accelerator

High-speed photography has been found to be an extremely valuable tool in connection with impact testing. With the aid of high-speed photographs we have been able to observe the production of fractures in the femur and of a fracture in the skull. The degree of deformation that occurs due to impact can also be readily observed with the aid of high-speed photography. Strain gauges, stresscoat, and photostress are used as tools in the observations of deformations and strains which occur due to impact. A solution of milling yellow, which is an analine dye, has proved very valuable for observing the presence of shear in a liquid medium subjected to impact, since this material is doubly refracting in the presence of shear stress. This double refraction can be observed and evaluated qualitatively in terms of intensity of the shear stress developed when viewed with the aid of polarized light. We also have a radioisotope laboratory available for use in our studies as occasion demands. Our most important asset, however, is the personnel in the various disciplines who work together on the problems that we undertake to study. These people come from the fields of anatomy, neurosurgery, pathology, engineering mechanics, electrical engineering, and, in fact, any discipline required for a particular investigation.

A procedure we are attempting to use whenever possible in the conduct of our investigations is as follows. Tests are conducted first on living animals. (Following the tests the animal is sacrificed and embalmed.) The second series of tests, identical to the first with regard to all the variables, is conducted on the embalmed animal, and the results are compared. Tests are then performed on human cadavers and, on the basis of the tests on living and dead animals, the data obtained from the human cadaver is extrapolated to the living. With the use of this technique we hope to achieve results as close as possible to those that would be obtained with the use of living humans as test subjects. One variable that we have not been able to take into account is the effect of the muscle force in the human. This is of course also not present in the anesthetized animal that we conduct tests on. That this force can be appreciable and have considerable effect on the results is well known. Clinical data are available indicating that bones in the body may be fractured by the application of internal muscle forces alone without the superposition of any externally applied force. This problem may ultimately be solved by tests made on volunteers from penal institutions.

Research regarding the effect of impact on the human body has barely begun. In head injury practically all the studies thus far involve impact by hard plane surfaces. The effect of impacts to various regions of the head by concave, convex, and relatively sharp surfaces needs investigation. We need to know the relation between the character of the impact and the resulting damage produced by blows to all locations of the human body. The Society of Automotive Engineers has established a committee for the determination of performance standards for padding and modification of the passenger compartments of automobiles to reduce injuries and fatalities resulting from crashes. Information is sorely needed to determine what degree of padding is necessary and what the characteristics of the padding should be to prevent injury to various parts of the body under various impact conditions.

A word of caution is in order regarding the method of describing the impact which results in injury of a specific severity. Goldman, von Gierke, and others have pointed out that the major portion of the body consists of material which is visco-elastic in character. As a first approximation, with small amplitude deformations the assumption that the material of the body behaves in a linear fashion is reasonable, and expressions for its behavior can be put in mathematical terms. However, when the deformations are large, as is the case in the production of injury, it is no longer possible to assume linear behavior of the material. When deformations large enough to produce injury occur, the visco-elastic characteristic of the material of the body must not be overlooked. It is therefore very important that the description of the impacts be adequate to take this property of the body into consideration. There has been a tendency in the past to assume that the impact was adequately described by the maximum value of the acceleration achieved. This procedure uses the magnitude of the acceleration multiplied by the weight of the body or some of its parts to describe a force which is assumed to be responsible for the damage. This technique is not even valid for rigid engineering materials, because the distribution of stress due to an impact is much different from the stress distribution resulting from a force. This is readily observed in engineering materials by the character of the failure of a part containing a stress concentration. Due to impact, the part fails at the point of stress concentration because of high-energy absorption in this region, but, with the application of a force, failure will generally not be influenced by a stress concentration. The behavior of any material under impact will be vastly different from its behavior under a static force, and to attempt to use the concept of force in describing the effects of an impact to visco-elastic material is entirely wrong. Any attempt to equate a dynamic problem in strength of materials to a static one

through the use of Newton's Law, F = Ma, will lead to fallacious results. Newton's Law was written only for a particle. When attempting to apply this law, F = Ma, to a visco-elastic material such as the body is composed of, and where a is some value read from an accelerometer attached to the body, the result obtained will have absolutely no significance with regard to the stresses produced in the body. With a blow to any visco-elastic material the time history of the event leading to the deformation and stress distribution in the material is of utmost importance. Recent tests of impacts to visco-elastic photoelastic materials by Dr. Paul D. Flynn of the Illinois Institute of Technology have provided a beautiful illustration of the many variables of the impact affecting the stresses set up in material of this type. His tests included both circular and rectangular models, and the impacts were produced by strikers of various weights dropped through various distances. In tests in which essentially the same energy was applied but strikers of different weights were used, the magnitudes of the maximum principal stresses were appreciably different, as were also the time histories and total durations of the stresses.

With regard to future impact research, we are currently designing a laboratory building specifically for biomechanics research at Wayne State University. This will include a vertical laboratory over 100 feet in height, to be used for drop testing and to contain an accelerator for producing controlled vertical accelerations. A long horizontal area will be provided for an accelerator which will have sufficient capacity to carry the body of an automobile to be oriented in any direction, in which cadavers can be subjected to programmed accelerations. A small centrifuge is planned for another part of the building. Other facilities will include tanks in which explosive-compression experiments may be conducted both in air and in water. An area will be provided to permit the 180-degree swing of a 40-foot pendulum in which a body may be mounted. Complete x-ray, radioisotope, and photographic laboratories will be provided as part of the facility, as well as necessary shops for the manufacture of instruments and equipment. An area in the basement will provide a noise- and vibration-free environment, while in another part of the building, vibration equipment will be used for testing the responses of humans and animals to vibrations. This facility is only in the planning stage, and no funds for its construction are as yet available.

A bibliography of the Biomechanics Research Center will be found in the Organization Index of the Chronological Bibliography at the end of this publication.

Institutions Indicating Interest in Conducting Impact and Vibration Research on Animals and Humans

Engineering Colleges

Agricultural and Mechanical College of Texas, College Station 4, Texas
University of Arkansas
University of California, Berkeley 4, California
University of California at Los Angeles, Los Angeles 24, California
Carnegie Institute of Technology, Pittsburgh 13, Pennsylvania
Columbia University, New York 27, New York
University of Denver, Denver, Colorado
Illinois Institute of Technology, Chicago 16, Illinois
Institute of Technology—Air University, Wright Patterson AF Base, Ohio
University of Kentucky, Lexington, Kentucky
Lafayette College, Easton, Pennsylvania
Manhattan College, New York

University of Maryland, College Park, Maryland
Massachusetts Institute of Technology, Cambridge 39, Massachusetts
University of Minnesota, Minneapolis, Minnesota
New Mexico State University, University Park, New Mexico
New York University, New York
Northwestern University, Evanston, Illinois
University of Toledo, Toledo, Ohio
Tuskegee Institute, Tuskegee, Alabama
Vanderbilt University, Nashville 5, Tennessee
Wayne State University, Detroit 2, Michigan
West Virginia University, Morgantown, West Virginia

College of Agriculture

University of Wisconsin (Department of Veterinary Science), Madison, Wisconsin

Medical Schools

University of California—San Francisco Medical Center, San Francisco, California
Duke University, Durham, North Carolina
University of Michigan, Ann Arbor, Michigan
Northwestern University, Evanston, Illinois
University of Texas, Galveston, Texas
University of Utah, Salt Lake City, Utah
Wayne State University, Detroit, Michigan

Department of Preventive Medicine

Ohio State University, Columbus 8, Ohio

Institute of Agricultural Medicine

State University of Iowa, Iowa City, Iowa

Note: Some of the institutions listed have expressed an interest in the field of ultrasonic research only.

AUTOMOTIVE IMPACT

A. L. Haynes
Director, Product Study Engineering Office
Ford Motor Company

Studies of automotive impacts have been undertaken in a research program to reduce human injuries in traffic accidents. Many aspects of accident circumstances have been investigated by means of at-the-scene observations, analysis of accident and injury reports and instrumented simulations of impact conditions. Detailed examinations of crashed cars have yielded important information on structural behavior under crash-impact loading. Medical reports frequently have indicated causative factors for occupant injuries. Statistical studies on traffic accidents have revealed cause-and-effect relationships in vehicle behavior and occupant-injury patterns which in many cases are not apparent during an investigation of a particular accident. However, a crash event occurs with such rapidity and transient forces are so variable that after-the-fact evaluations of actual traffic accidents are more speculative than analytic.

The problem was to explore and define the mechanisms operating during a collision, to determine the characteristics of the forces generated, to measure the loading imposed on human occupants, and to establish the human tolerance to such loading. It appeared logical that, guided by factual knowledge of what actually transpired in an automotive collision and how the occupants responded to force applications, vehicle designs could be modified and special features could be developed which would reduce the injury potential in automotive accidents.

The immediate task was to synthesize model crash conditions which would be realistic in terms of actual traffic experience and at the same time be reasonably controllable. A survey of accidental collision circumstances demonstrated a multitude of factors were involved in individual accidents. It obviously was impracticable to attempt simulations of all the various accidents which could occur. Accident-case analysis showed that the majority of traffic accidents were car-to-car and non-collision, single-car involvements. However, these were the particular accident types which exhibited the widest range in vehicle damage and occupant injuries. On the other hand, nearly 20 per cent of accidents were found to be impacts into unyielding objects such as bridge abutments, large trees, and solid embankments. Such crashes were shown to result in substantially similar vehicle-deformation and occupant-injury patterns for comparable impact speeds.

The indicated reproducibility of barrier-type collisions recommended this configuration for obtaining data on the effects of impact speed, vehicle weight, and structural design. Staging a barrier crash under controlled conditions had the added advantage of locating the total event within a confined area. Consequently, high-speed photography of the collision could be facilitated with fewer problems involved in recording the outputs of electronic instrumentation installed on the vehicle and simulated occupants.

Fully instrumented car-to-car collisions, usually of a broadside configuration which occurs frequently in urban-intersection accidents, also are conducted under controlled conditions, and rollover studies are run to observe vehicle and occupant behavior in this type of accident. As anticipated, data from the car-to-car crashes have been found less consistent than the results obtained from barrier impacts. Lack of reproducibility particularly is observed in the higher speed ranges where extensive mutual penetration and displacement of the colliding vehicles occur. Behavior patterns also are influenced by differences in vehicle weight, variations in structural design, and in the vehicle areas which come into contact during the crash. However, when similar cars are crashed under controlled conditions, a plot of passenger-compartment peak deceleration-versus-impact speed for car-to-car collisions provides a curve which can be compared with similar data from barrier impacts to demonstrate the relative force severity for these two types of crash circumstances.

Figure 1 compares the peak decelerations measured in the passenger compartments during barrier and car-to-car collisions of similar cars. In the lower-speed impacts, the peak deceleration attained during a barrier crash is approximately twice

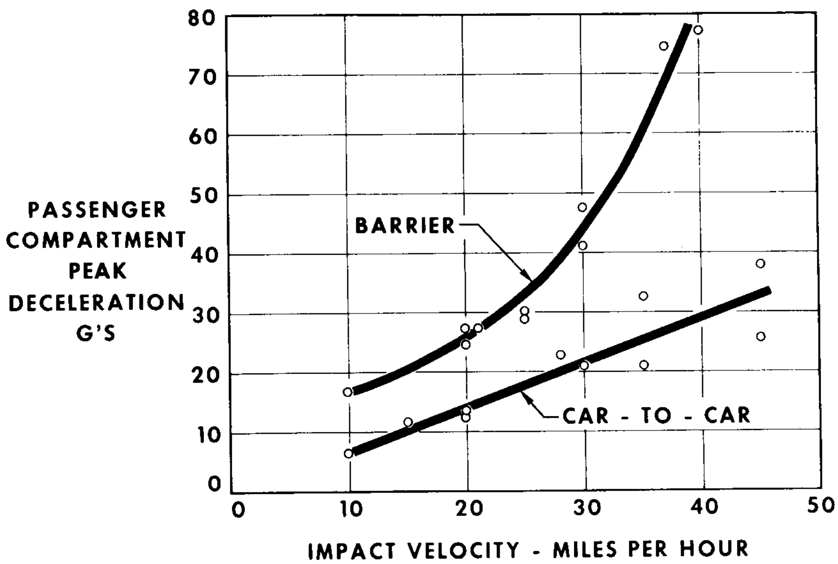


Figure 1. Deceleration Comparison—Car-to-Car and Barrier Crashes.

the level measured in a car-to-car collision. With higher-impact speeds the deceleration ratio progressively increases. The spread in the car-to-car data is apparent, while the barrier measurements show good reproducibility. Because consistent data are provided by barrier impacts, such crashes can be used as a reference when comparing the effects of design changes and determining the energy-absorbing properties of special features which are developed for reducing potential occupant injuries in collisions.

In some cases, it is possible to compare the structural damage observed in an actual accident with that obtained in a research crash test. In these cases, the dynamic loading data measured during the crash study permits an approximation of the loading to which human occupants were exposed during the traffic accident. Correlation of the injuries received by these occupants with the probable loads imposed provides an indication of human force tolerance.

Figure 2 is a schematic of the set-up used for conducting barrier-crash studies. A Thunderbird is used as the towing vehicle and attaches to a bridle on the test car by means of a quick-disconnect cable. No brakes are applied during a barrier crash, so the test car moves into the barrier under free momentum. All the instrumentation

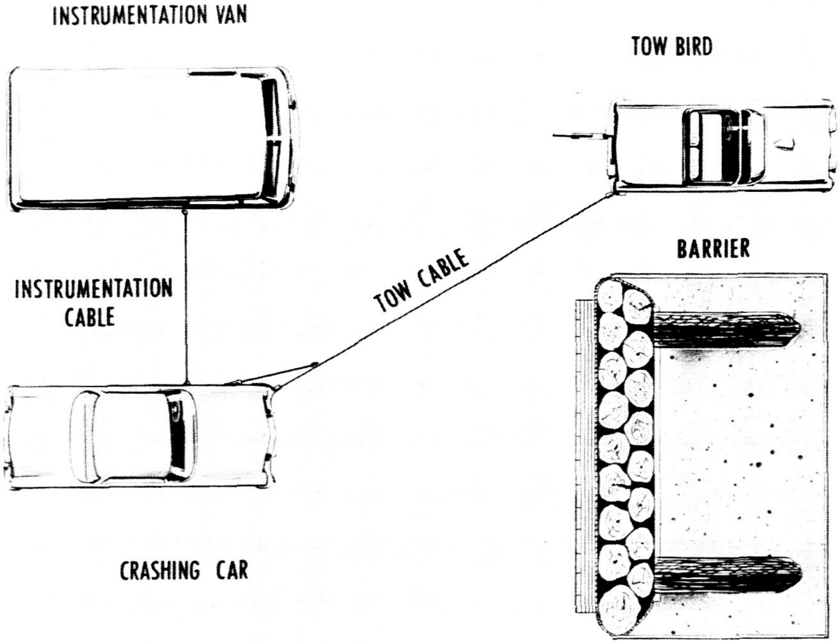


Figure 2. Barrier Crash Schematic.

gear, in the crashing car and on the dummy occupants, is coupled to the instrumentation van by a multi-wire cable supported by a shock cord. The instrumentation van supplies the power for the electronic instrumentation and carries the high-speed, multi-channel recorder which provides the tracings of the transducers in the crashing car. Gunsight motion-picture cameras are carried on the crashing car to record occupant kinematics. Various high-speed motion-picture cameras also are disposed about the site of the impact to record the collision events. Targets on the vehicle and dummies and affixed to the roadway provide markings for motion analysis of the film records. A flash bulb, located on the crashing car, is actuated by a contact switch on the front bumper to indicate when the barrier is contacted.

Figure 3 illustrates the construction details of the eighteen-foot-wide barrier. Two-foot-diameter logs, twelve feet long, are embedded vertically six feet in the ground. A one-half-inch-diameter steel cable ties the logs together near the top, and a [illegible]-inches-thick slab of concrete binds them just below the surface of the ground. The logs are backed by sand fill retained by planking. The front face of the barrier is provided with replaceable oak boards, which are supported in the crevices with concrete fill. The rigidity and durability of the barrier permits repeated crashes with identical configurations.

The crash-test dummies, shown in Figure 4, are fully articulated to simulate the skeletal flexibility of the humans they represent. This design feature permits a realistic evaluation of upper-torso jackknifing and the flailing of extremities that occur in a crash with human occupants. The weight and weight distribution of these

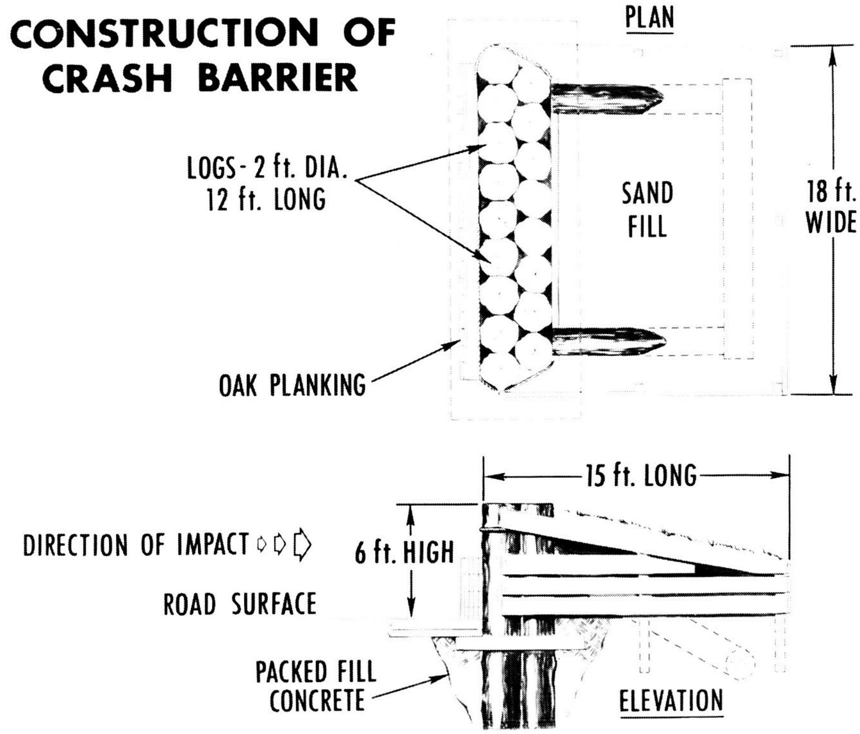


Figure 3. Research Crash Barrier.

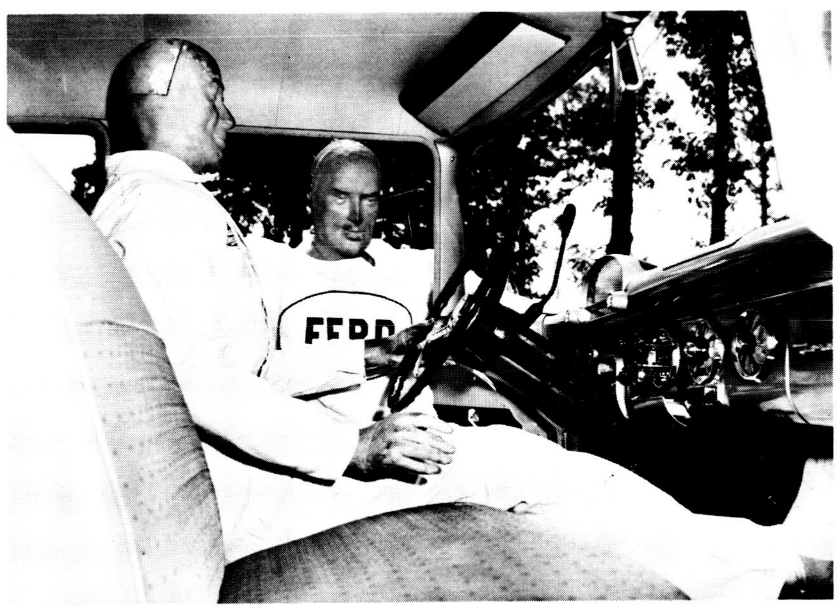


Figure 4. Anthropomorphic Test Dummies.

anthropomorphic dummies are comparable to those of an adult human. Total dummy weights, for the various models used, vary from 171 to 217 pounds. Overall heights range from five feet nine inches to six feet, and seated heights are from 35-1/2 inches to 40 inches. The rear quadrant of the head is removable, permitting the installation of accelerometers on the aluminum head casting, and cavities in the chest and abdomen

can be fitted with load transducers. The surface layer of the dummies is a flexible plastic material which simulates human soft tissue in elastic response. To determine areas contacted by the dummies during a collision, aluminum foil can be attached over the forehead, knees, or other areas to form one contact of a switch, while a similar foil application on interior surfaces of the vehicle provides the other contact for the circuit. By this means, the force measurements from the dummy transducers can be related to the areas contacted.

Figure 5 illustrates the installation of a bar-slide tensiometer on one tail of the seat-belt webbing. Stress set up in the webbing, as the seat belt is loaded during a

Figure 5. Belt Tensiometer Installation.

collision, causes a strain curve in the instrumentation records from which peak loading, rate of onset, and loading duration can be determined. The sum of the strains developed in both sections of webbing encircling the pelvic area of the dummy totals the loading imposed in the seat belt during the impact. Similar strain gauges also can be installed on other structural elements, such as the door-latch components, to determine the stress developed during a collision impact.

Figure 6 is a view of the chassis showing the installation of an accelerometer on a frame rail. For unitized structures, this installation is made to welded-on brackets to the floor pan and the stub frame rails in the engine compartment. Accelerometers are mounted at various locations on the car structure to determine the absolute and relative decelerations of the vehicle. For establishing the relationship of passenger-compartment deceleration and seat-belt loading, the tracings from the accelerometers located in the area of seat-belt floor anchors are used.

Figure 7 presents barrier-crash pictures which illustrate exterior damage and interior views into a 1955 Mercury and a 1958 Lincoln. The Mercury, which struck the barrier at 27 mph, employed frame-body construction. The Lincoln, with unitized structure, impacted the barrier at 29 mph. The total deformation of the Mercury structure was greater than that of the Lincoln. However, the Lincoln deformation

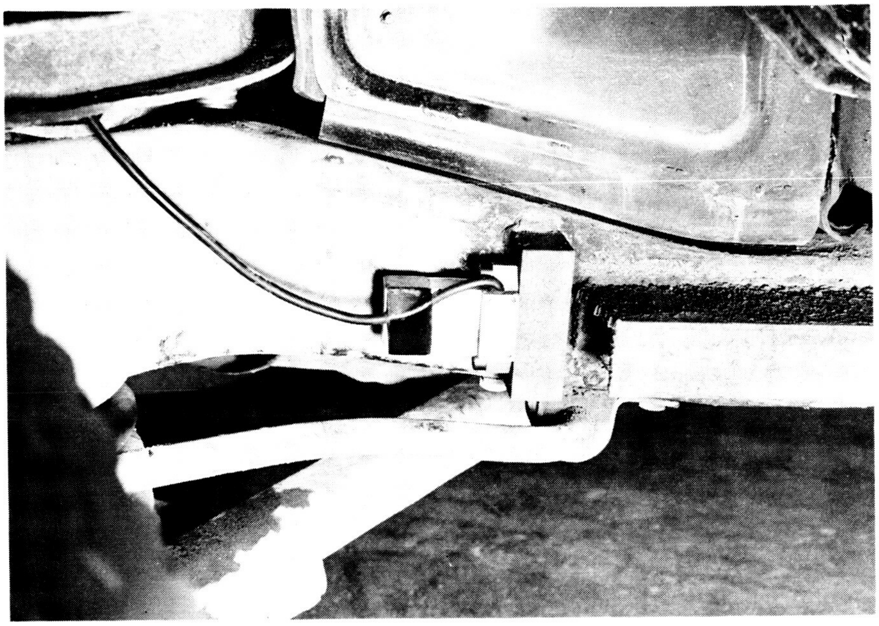


Figure 6. Chassis Accelerometer Installation.

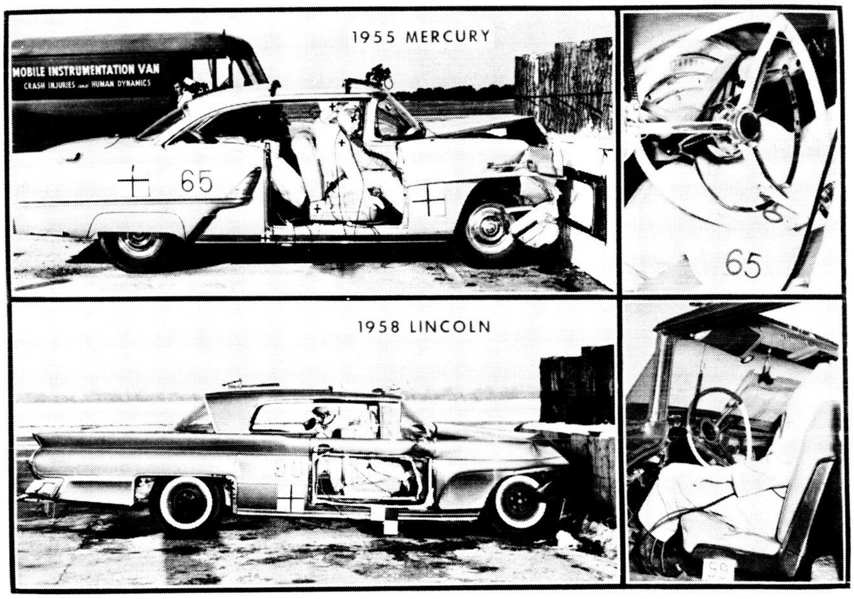


Figure 7. Barrier Crashes.

extended for a greater length in the structure as witnessed by the collapse in the r quarter panel just forward of and above the rear wheel. Since the primary structure in the Mercury was the frame, it was possible to remove the right front door for photographic purposes without materially affecting structural deformation characteristics. Because of the unitized structure in the Lincoln, only the panels of the door were removed and the longitudinal door structure was retained to maintain overall vehicle stiffness.

The driver dummy in the Mercury wore a seat belt which kept his knees from contacting the instrument panel. Without a seat belt in the Lincoln, the knees of the driver dented the instrument panel and his head struck the padded sun visor. Door areas were cleared in these cars to permit high-speed motion pictures of occupant and seat movements. The right front seat occupants in both cars wore seat belts, and their crash kinematics were related to recordings of seat-belt loadings.

Although the Mercury was of 1955 vintage, a recessed-hub steering wheel was installed, double-grip door latches were incorporated, and padded instrument panels and sun visors were used to provide protective features comparable to those in the 1958 Lincoln. These safety features have been available in all Ford-built cars beginning with the introduction of the 1956 models.

Selected frames from high-speed movies of a 25-mph barrier crash are shown in Figure 8. The time in seconds after barrier contact is noted on each frame sequence.

Figure 8. Barrier Crash Kinematics.

Collapse of the front-end structures progressed for the first 60 milliseconds. Seat and occupant movement did not begin until practically all the vehicle deformation had taken place. At shortly after 60 milliseconds, forward velocity of the vehicle ceased. At this time the occupant was sitting vertically upright in the seat and peak seat-belt loading was attained. By 80 milliseconds, the front-seat passenger had begun to jackknife over the seat belt, and the unrestrained rear-seat passenger was moving forward. The advance of the rear occupant continued as his head struck and dented the roof panel. It is interesting to note that the rear occupant continued to travel forward after the front-seat occupant had completed jackknifing and was being drawn back into his seat by restitution in the seat-belt loop encircling his abdomen. In less than three-quarters of a second after impact with the barrier, the rear passenger had completed his forward travel and returned to the rear seat.

Figure 9 graphically illustrates vehicle deformation and load data for the barrier crashes of the Mercury and Lincoln shown previously (Figure 7). Although the motion-picture frame sequences shown in Figure 8 were for a slightly lower-speed barrier impact of a different car, similar events occurred on a slightly extended time scale during the Mercury and Lincoln crashes.

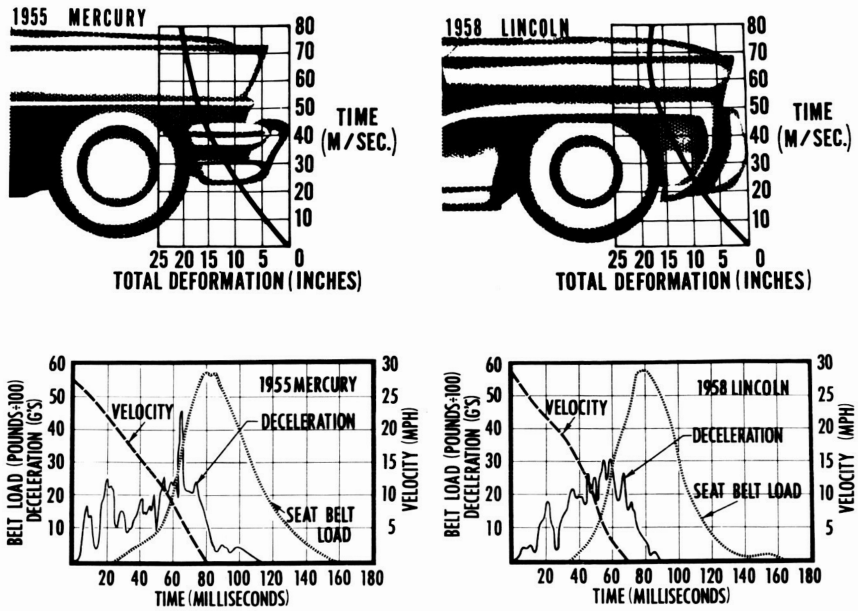


Figure 9. Deformation, Velocity, Deceleration and Belt Loading.

Plotted over outlines of front-end structure are the deformation-time curves for vehicle collapse. The lower graphs plot the passenger-compartment deceleration and velocity and the seat-belt loading pattern as a function of time after barrier contact. It will be noted that the vehicles were stopped in 70 to 80 milliseconds. Peak seat-belt loading was attained at this time. As previously illustrated (Figure 8), the belt-wearing front-seat passenger was sitting vertically upright when forward motion of the vehicle ceased. As the occupant jackknifed forward over the seat belt, pivoting at the hip joints, the seat-belt load began to fall and the hips were drawn rearward in the seat. As a result of this hip movement, the spine of the occupant was drawn axially rearward as the upper torso arched forward. Consequently, in the case of the Mercury, the forehead of the dummy received a glancing blow as the padded instrument panel was struck. In the Lincoln, which had greater front-passenger-compartment clearance, the occupant jackknifed full forward without striking any interior structure.

The passenger-compartment deceleration traces show a series of shock loadings, which occurred when structural components dumped energy as the front end of the vehicle progressively collapsed. The rigid bumper and frame structure of the Mercury gave rise to an abrupt initial deceleration impulse, while the deceleration of the Lincoln structure produced less pronounced peaks as the vehicle was brought to rest. It is instructive to observe that the shock-loading pattern of the passenger-compartment deceleration was not reflected in the smooth rise and fall in seat-belt loading. This was due to the damping effects of the seat-belt system. It is particularly significant

to note that, although these representative barrier crashes were severe impacts which produced almost three tons seat-belt loading, the belt-loading patterns were essentially triangular and indicated that considerable energy was absorbed as the vehicle structure was deformed forward of the passenger compartment.

Figure 10 provides a better perspective of the dynamic patterns which result when similar cars are tested in barrier impacts and broadside car-to-car collisions.

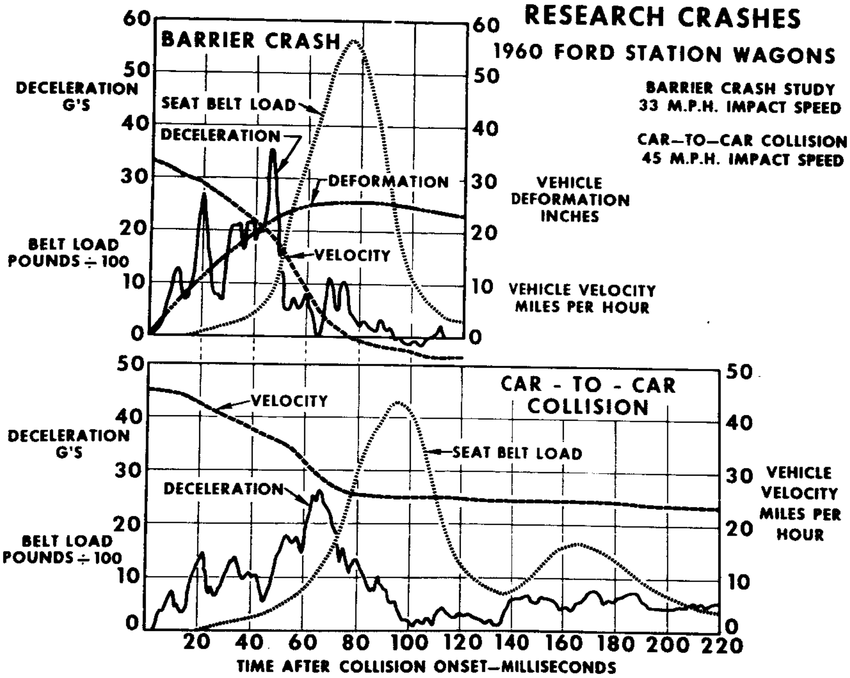
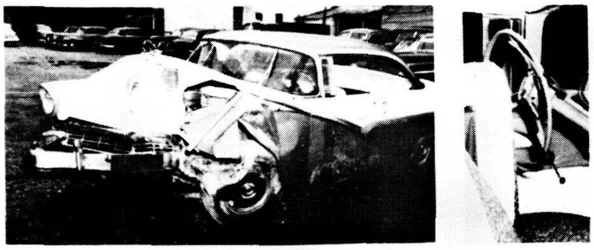


Figure 10. Barrier and Car-to-Car Performance Comparisons.

These data were obtained during research crashes of 1960 Ford station wagons. The barrier impact was at 33 mph, and the car-to-car collision was at 45 mph. Passenger-compartment peak decelerations of 36 G's and 26 G's respectively were measured during these crashes. These values, which show a 10-G higher deceleration for a barrier impact occurring at 12 miles an hour slower than the car-to-car collision, are consistent with the relative data plotted in Figure 1. The extensive mutual penetration of the colliding cars and displacement of the struck car in the car-to-car collison delayed the build-up to peak deceleration and seat-belt loading. The deceleration impulses also are less pronounced during the car-to-car crash, due to the damping effects of yielding structure. In the barrier impact, forward velocity reached zero in less than 100 milliseconds, while the striking-car velocity was still 23 mph at the end of 220 milliseconds in the car-to-car collision. The effects of the relative speed change in the two vehicles is reflected in the comparative seat-belt loading patterns. The belt loading in the car-to-car collision attained a lower peak value than in the barrier crash, but persisted for a longer period. A characteristic second rise in deceleration amplitude, followed by a rise and fall in seat-belt loading, is displayed in the car-to-car collision curves. From high-speed motion-picture analysis, it is apparent this phenomenon is caused by bottoming out of the initial structure penetration and an increased resistance to further deformation while the struck car is moved sideways by the impacting vehicle.

In Figure 11, the 33-mph barrier crash, referred to in the previous graphs (Figure 10), is compared to an actual traffic collision with a 10-inch-diameter tree.



DRIVER SUSTAINED MINIMAL CHEST BRUISING

ACTUAL ACCIDENT - CAR STRUCK 10 INCH DIA. TREE

RESEARCH CRASH STUDY - BARRIER IMPACTED AT 33 M.P.H.

Figure 11. Accident and Crash Study Comparison.

The exterior damage resulting from both collisions was substantially similar for the two cars. The deformation of the steering wheels also was similar. Since driver weight was almost the same in both crashes, it is probable that almost equal loads were supported by the chests of the human driver and the anthropomorphic dummy driver to result in the same degree of steering-wheel deformation. The medical report on the actual accident listed only minor bruising to the driver's chest. Therefore, it can be inferred that an energy-absorbing steering wheel can be collapsed to this degree while causing only a minimal chest injury.

In Figure 12, an actual traffic car-to-car collision is compared to the 45-mph car-to-car collision of the station wagon referred to in Figure 10. The similarity of front-end damage and steering-wheel deformation is apparent. From the pictorial similarity of the two collisions, it can be assumed that approximately the same loading conditions were experienced in both crashes. Since no chest injury was reported for the driver in the traffic accident, chest-impact loadings which produce such minor steering-wheel deformation can be considered well within the range of human tolerance.

Figure 13 illustrates a steering-wheel-impact fixture used for laboratory studies. It is difficult to obtain reliable load measurements for chest impacts against steering wheels during full-scale crash tests. This laboratory impact technique was developed to provide realistic simulations of the conditions found to exist in actual collisions. A torso-shaped mass is used as a pendulum for impacting the steering wheel. The steering wheel is mounted at the car-installation angle by means of a nose piece on a horizontal load cell. The impact torso weighs 150 pounds and is hinged to the pendulum shaft at the level of the hip joints. A shear-pin, installed on the rear of the head, permits the torso to jackknife forward and bring the chest against the steering-wheel rim after the abdomen has first made contact. Micro-motion studies of high-speed

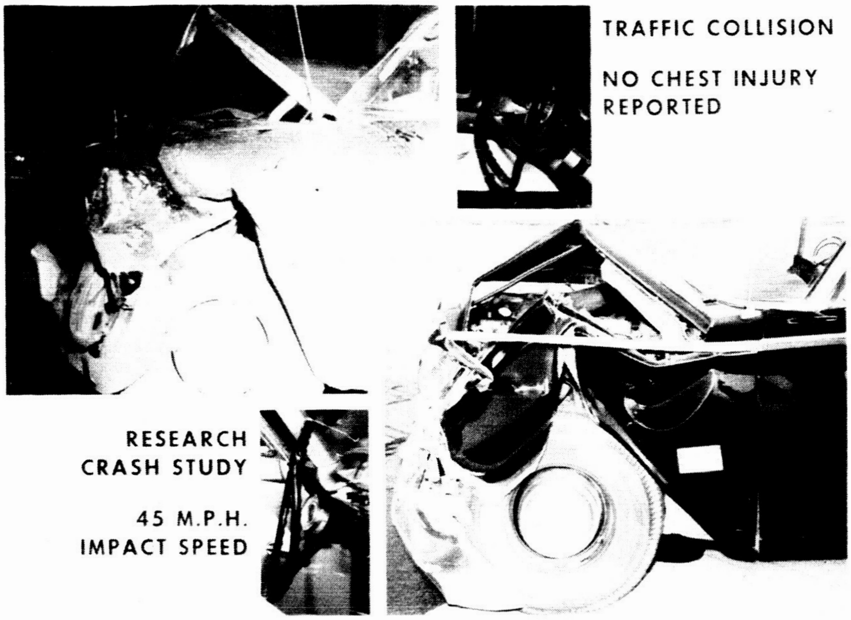


Figure 12. Accident and Crash Study Comparison.

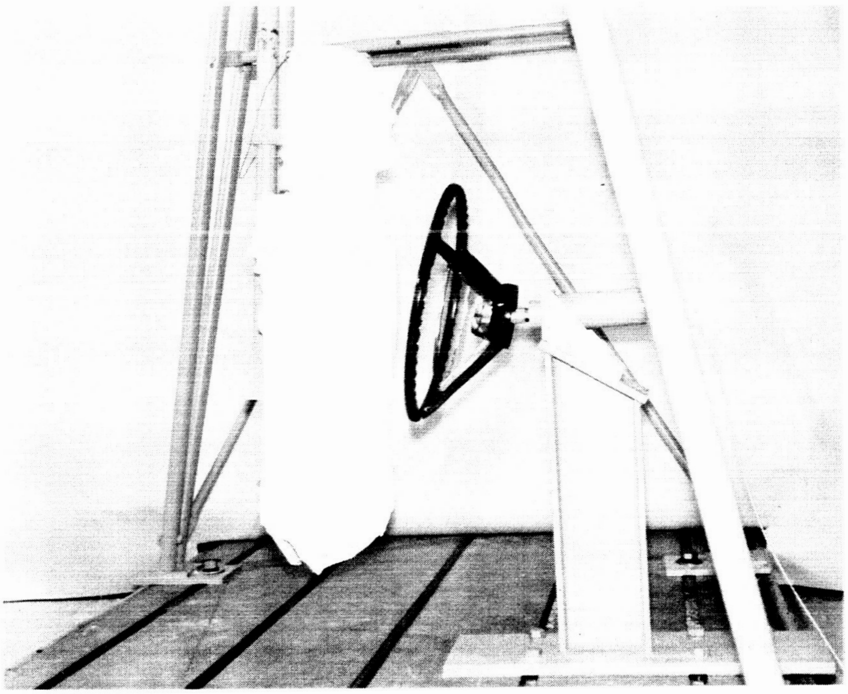


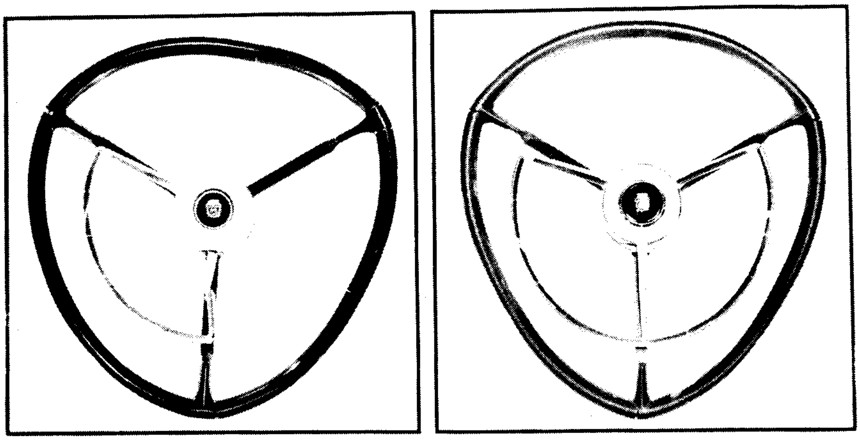
Figure 13. Laboratory Impact Fixture.

motion-picture film demonstrated these to be the typical kinematics of a driver impacting a steering wheel during research full-scale crash studies.

A method also has been developed for calculating pressure-time data during the laboratory steering-wheel impact. The wheel rim is sprayed with paint and the impact torso, with a clean cloth applied over the front surface, is dropped against the wheel

before the paint has dried. High-speed motion pictures of the impact show the body area progressively making contact as the steering-wheel rim is deformed. By comparing the load measurements and contact areas, a pressure-time curve can be plotted for both the abdomen and chest impact.

Figure 14 illustrates how the deformation of a steering wheel in a laboratory impact study compares with the wheel damage obtained in a 45-mph car-to-car collision



DRIVER CHEST IMPACT

45 MILE PER HOUR
CAR-TO-CAR CRASH STUDY

TORSO BLOCK IMPACT

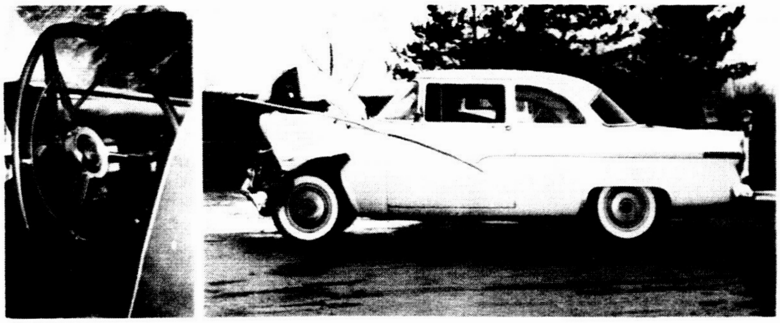
15 MILE PER HOUR
LABORATORY TEST

Figure 14. Steering Wheel Deformation Comparison.

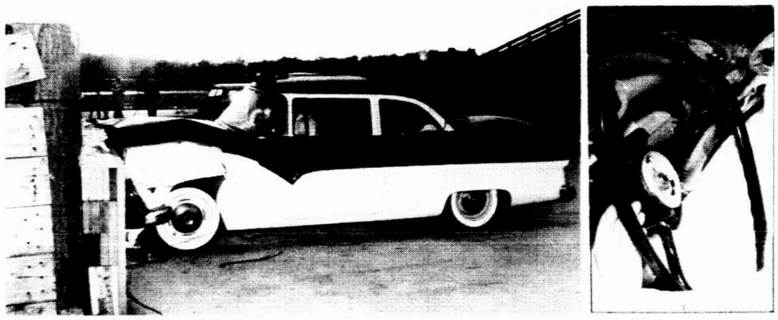
study. The steering-wheel deformation is almost identical when the impact pendulum struck the wheel at 15 mph. By comparing wheels impacted in the laboratory at different speeds to those deformed to the same degree in vehicle collisions, it is possible to estimate the impact load imposed on the driver's chest. Such data then can be related to the medical reports on the injury sustained during actual accidents to provide an approximation of human force tolerance.

Figure 15 compares a highway collision with a stationary truck and a 25-mph barrier-crash study. There was a similar degree of exterior damage and steering-wheel deformation in both crashes. It will be observed that, although only minor steering-wheel deformation has occurred in these relatively severe collisions, the drivers in both cars were wearing seat belts. The seat belt supported a substantial portion of the driver's weight, and thereby reduced the loading imposed on his chest during the steering-wheel impact.

Figure 16 shows a micro-motion study of the kinematics observed for a righ front-seat passenger during a 20-mph barrier impact when seat belts were not worn. The occupant's knees struck and dented the face of the instrument panel. The upper torso then travelled forward. The chest struck and severely deformed the upper crown of the instrument panel, and the head impacted and penetrated the windshield. A still photograph of the actual interior damage illustrates the high injury potential when the passenger continues to travel forward or the vehicle is stopped abruptly by the barrier.



ACTUAL ACCIDENT - CAR STRUCK STATIONARY TRUCK



RESEARCH CRASH STUDY - BARRIER IMPACTED AT 25 M.P.H.

Figure 15. Accident and Crash Study Comparison—Seat Belts Worn by Drivers.

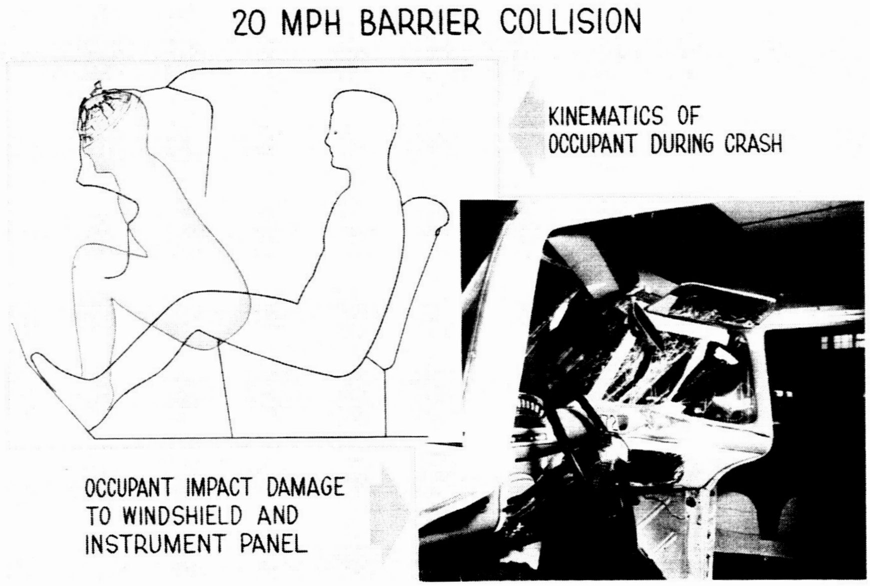


Figure 16. Occupant Without Seat Belt in Barrier Crash.

Figure 17 illustrates how the occupant is retarded with the vehicle when seat belts are worn. As previously illustrated (Figure 8), the passenger jackknifes forward on the belt and the forehead strikes the surface of the instrument panel with a glancing blow, since the head is moving rearward at the moment of contact. Although this barrier crash occurred at 25 mph, while the impact without seat belts was at only 20 mph, the photograph illustrates the substantial reduction in injury potential when seat belts are worn.

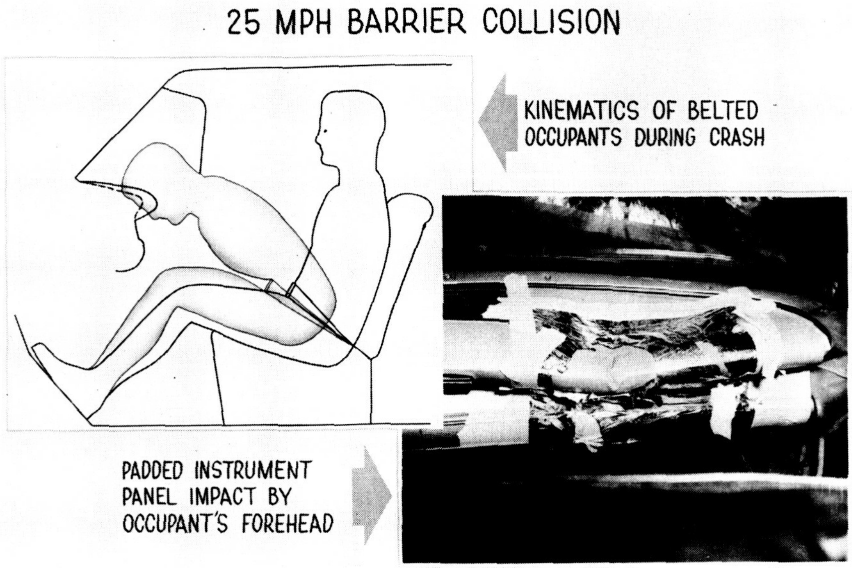


Figure 17. Occupant Wearing Seat Belt in Barrier Crash.

Since the seat belt can control the path taken by the upper torso as jackknifing occurs, it is practical to apply energy-absorbing padding over structure surfaces that can be contacted. Properly compounded padding material with adequate thickness will distribute the impact energy over a substantial body area and greatly reduce the potential injury sustained when interior structure is struck by the occupant.

Some of the techniques employed for obtaining factual data on automotive impacts have been outlined briefly in this paper. These research studies have been conducted as part of an ever-expanding program to develop design features and methods for reducing occupant injuries in actual accidents. Some values for human tolerance to impact forces have been estimated by the comparisons of actual accidents and occupant injuries with the force and kinematic data obtained in research impact studies. Techniques have been developed for predicting the loading patterns and structural behavior characteristics which can be anticipated under particular crash-impact conditions. However, much less is known about the human response to impact loading. Studies in this important field are underway at various medical centers, frequently with the financial sponsorship of automobile manufacturers. Statistical analysis of accidents involving late model cars have demonstrated that a reduction in occupant injuries already has been achieved. It is anticipated that cooperative vehicle-safety programs now operating will result in considerable further progress in improving automotive transportation safety.

IMPACT STUDIES OF THE UNITED STATES AEROSPACE INDUSTRY

Galen A. Holcomb
Stanley Aviation Corporation

Introduction

A brief survey of recent impact studies by the United States aerospace industry reveals some rather singular achievements. With few exceptions, these studies have been the direct result of a need to determine the tolerability of a particular man-machine system under a given set of specification conditions.

Few of these studies have been experimental in nature where data on human tolerance to impact acceleration was derived. In fact, most have been confined to a literature search in which all of the known impact-acceleration-tolerance data were assembled and used as a guide in the interpretation of a rigorous, but confusing, specification human tolerance allowable during design. Others have engaged in analytical studies to determine the meaningfulness of this data. More precisely stated, this has consisted of scrutiny of the experimental test conditions to determine the validity of an accepted test point. This has also included studies to determine their statistical significance.

Other important analytical studies have been directed towards the development of mathematical techniques similar to those developed in other sciences, so that an assessment of the system capability can be made early in design before fabrication using time-saving devices such as computers, rather than relying mostly on test data after fabrication, as is now the case.

The importance of this approach cannot be overemphasized, since it has been the writer's observation that the lack of applied dynamics and mathematics by competent personnel to the field of impact-acceleration stress has been a major factor contributing to engineering waste during design.

Finally, a very few have engaged in actual biological experiments, the data from which might be classed as either applied or basic research. In most cases, however, these tests arose from the necessity to test a particular system rather than from industry's desire to engage in basic research.

This paper will discuss in greater detail the problems facing industry in the design of man-machine systems subjected to impact-acceleration stress, and what research industry has been doing and is doing in connection with these problems.

Influence on Design—A Statement of the Problem

It may appear to some that, in view of the extensive impact-acceleration research accomplished by some early investigators such as Stapp[6], Hessberg, Mosely,

Beeding[7], et al, and the demonstrated ability of Lombard[8] to assemble impact-acceleration-tolerance points by analysis and extrapolation, there is little need for industry to engage in further impact research. Needless to say, their work speaks for itself and was accomplished with great dedication and courage. However, important ingredients are still missing from published impact-acceleration-tolerance data making it limited in its use to the designer and industry as applied to a practical design problem. These are: (1) lack of extremely abrupt acceleration-tolerance data, (2) lack of data showing the effects of abrupt pulses or "spikes" superimposed on sustained pulses, and (3) lack of a well defined mathematical analogy and methodology understandable to the engineering profession.

Today's designer in industry, bogged down with such things as the definition of pulse "duration" and the tolerability or intolerability of the pulse slope, finally turns to the works of Kornhauser[10, 11, 22], Payne[9, 14, 15, 16, 17], Brock[12], and Shapland[18, 21], all competent dynamicists in industry, in an attempt to make mathematical order out of chaos. Inserted into their mathematics, of course, are the significant findings of von Gierke, Coerman, et al ASD[23], from their splendid research in vibration. These studies are in their infancy, however, and the results are found in few, if any, specifications. Thus, the use of the mathematical approach must be on an informal basis at this time and is subject to scrutiny and scepticism by some medical scientists whose job it is to pass judgment on the tolerability of a system as the design progresses.

In industry, the design of a complex aero-weapons system is accomplished by the melding of many disciplines into a team effort, each with well-defined areas of responsibility. The human engineers usually start with a task analysis followed by space layouts. After the weight engineers, structural engineers, and aerodynamicists have defined a preliminary exterior shape, the aerodynamicist takes over the chief role, for it is his responsibility to determine the lift and drag characteristics necessary to insure specification flight performance. In addition, he often finds it necessary to incorporate drag devices of some type which may be deployed during periods of high-impact pressure, in turn causing abrupt impact acceleration on the occupant.

To those who may not be familiar with engineering procedure, it should be noted that engineering design does not take place in a logical step-by-step fashion, solving one problem in an orderly manner before proceeding to the next. If that were the procedure, we might never finish, for the system would no doubt become obsolete by that time.

Actually, we proceed en masse after the preliminary design is completed, and the system is subjected to almost daily compromise between the weight engineers and the structures engineer, the human and the systems engineer, the aerodynamicist and the thermo-dynamicist, ad infinitum. Even as these struggles go on, drawings are being released and parts are being manufactured with the full realization that they may be changed or scrapped the following day. Decisions and compromises are made by means of daily conferences with representation from each area of responsibility. Most have handbooks to which they can refer for basic information. For example, the materials engineer uses a Rockwell hardness index, the structures engineer could not function without his ANC-5 manual, etc.

Imagine then, if you will, one of these conferences in which decisions must be reached. The aerodynamics department has just received an acceleration trace of the predicted flight performance of a system. They are asking the life sciences or human

factors department if these accelerations are tolerable to the occupant and if they meet the specification requirement. If they are not, new methods of aerodynamic control must be designed and wind-tunnel tested, causing delays and possible degradation of the total system due to weight and reliability compromise. The design department must cancel many drawings and begin new designs; production must be re-scheduled and most likely many parts scrapped; certain tests are made obsolete and new tests must be scheduled; and, last but not least, the contracting and procurement agencies must be briefed as to the reasons for delays and overruns.

At the conference, a decision has been requested and a decision will be reached for engineering design can proceed in no other way. Design decisions are made on the basis of the best information available on that day.

Figure 1 and Figure 2 might be considered typical traces to be analyzed for tolerability, the first being from flight accelerations and the second from ground impact. In both cases, the magnitude exceeds 35 G's and the slope exceeds 1,500 G/sec, representing the maximum allowable limits along the body X axis shown in the system specification. These limits, shown in Figure 3 are derived from HIAD, and are usually accompanied by no explanation as to their intended use or meaning.

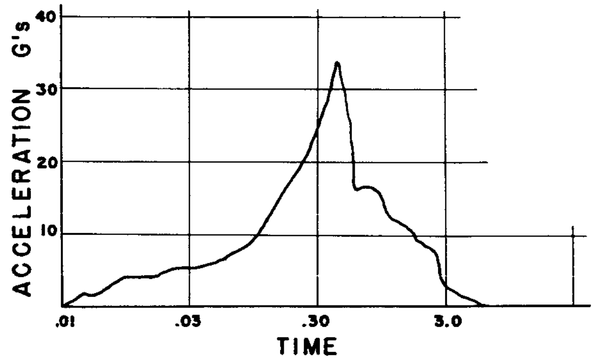


Figure 1. Typical acceleration trace during flight conditions.

The writer has previously discussed the shortcomings of HIAD[1, 2, 24] tolerance data in detail, and will not devote further time to the subject in this paper, except to summarize. This is necessary at this time in order to demonstrate that, in the impact regime, industry has had no choice but to engage in impact studies, since it is usually impossible to arrive at design decisions using the information available in specifications and/or HIAD[24]. If industry were to apply these allowables literally, more often than not (as has happened in the past) system design would be unnecessarily compromised and costly.

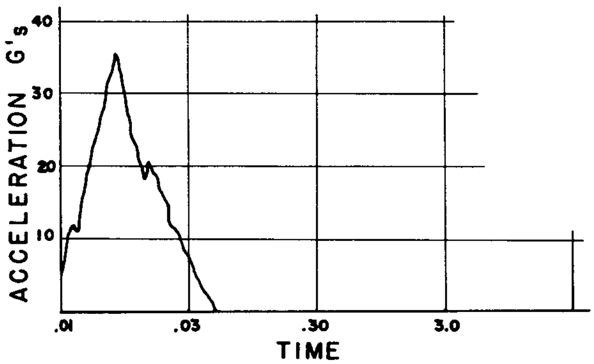


Figure 2. Typical acceleration trace during ground impact.

Although this graph (Figure 3) filled a void at the time it was first introduced into HIAD[24], it is now a serious impedance in the design circuitry. Some of the reasons are:

1. Since the term "time" is used along the abscissa instead of "duration," and a rate-of-onset slope is contained at its beginning, a time-G history is inferred. Examples are plentiful, including Government documents, in which acceleration traces are matched against this graph to assess tolerability. Actually, assessment in this manner has no meaning, and the area under this curve has no significance. Figure 4 illustrates.

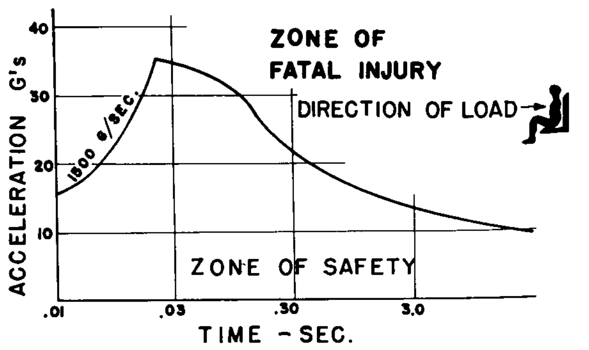


Figure 3. Graph showing human tolerance to acceleration along the body X axis as contained in HIAD(24).

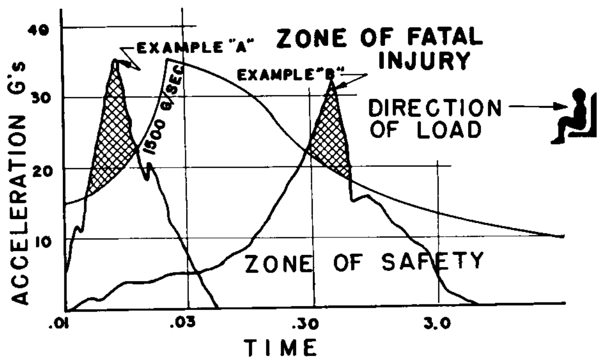


Figure 4. Typical example of matching Time-G history to a Duration vs. G Graph. This method is incorrect.

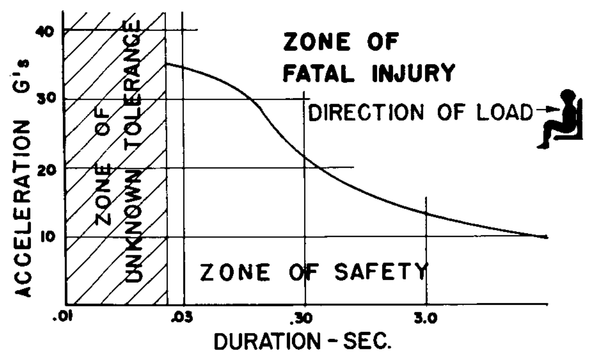


Figure 5. Corrected graph showing void in tolerance data durations less than .023 seconds.

2. The rate-of-onset slope is out of context as a part of this graph. Since the graph is a series of relationships between G magnitude and duration, this relationship is not valid along this slope. Therefore, with its removal, it can be seen that the graph is lacking test points with plateau durations less than .023 seconds (Figure 5).

3. As reported by Eiband(3) and others(1, 2), the data from which this graph (Figure 3) was evolved was largely based on G plateaus calculated from test vehicles. Since there are no test points on the graph with plateau durations less than .023 seconds (Figure 6), it is not possible to assess the tolerability of extremely abrupt pulses, such as ground impact, since they are essentially triangular in shape (Figure 7). In addition, within an extremely abrupt impact regime, the slopes exceed 1,500 G/sec. These have been shown to be tolerable, both analytically and experimentally(1, 2, 9, 11).

The purpose of this summarization is to illustrate what the inevitable decision must be at this engineering conference, re-enacted almost daily in industry, if the specification and/or HIAD(24) were followed literally. The design would be rejected on at least two points, i.e., (1) the rate-of-onset exceeds 1,500 G/sec, and (2) the magnitude exceeds 35 G's.

The results of this type of decision are tragic, and the costs to our defense efforts in time and money have been appalling. The aerospace industry has been placed in an untenable position by this state of affairs. Our very reputations are at stake, and the successful completion of contracts taken in good faith have been seriously jeopardized. As a result, the aerospace industry has served notice it intends to engage this problem in full concert with the services and universities.

Analytical Studies

It is the writer's opinion, shared generally in the aerospace industry, that impact acceleration is a dynamic problem that cannot be solved solely by the medical

profession. Neither can it be solved solely by the engineers. Great accomplishments can be achieved only by competent teams representing several disciplines, the most important of which are the medical sciences, mathematical dynamics, and engineering.

There have been cases in which engineers have attempted to do acceleration research delving dangerously into medical areas. However, there have been, perhaps, more cases of medical scientists doing impact-acceleration research delving equally dangerously into the field of dynamics without adequate support or training.

Early in 1961, Stanley Aviation, under NASA Contract NASr-37, set forth to review all of the existing physiological data dealing with abrupt acceleration. The goal of this program was to formulate mathematical models and analogies dealing with the responses of human subjects when subjected to acceleration in various restraint systems. After perusing most of the available data, it was found that very few experiments have been made in which the instrumentation and test conditions were such that valid mathematical answers could be obtained.

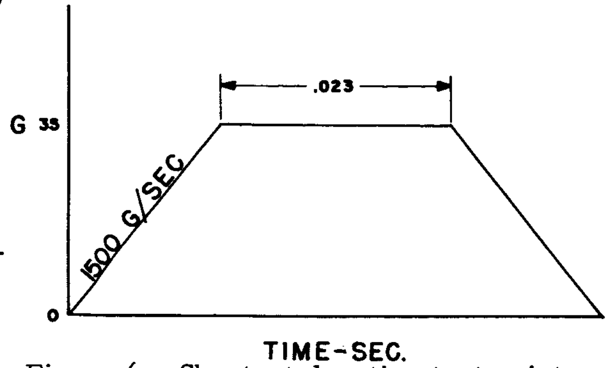


Figure 6. Shortest duration test point shown on the HIAD graph. "Duration" refers to plateau. Assessment of shorter duration plateaus or triangular pulses not possible per Figure 7.

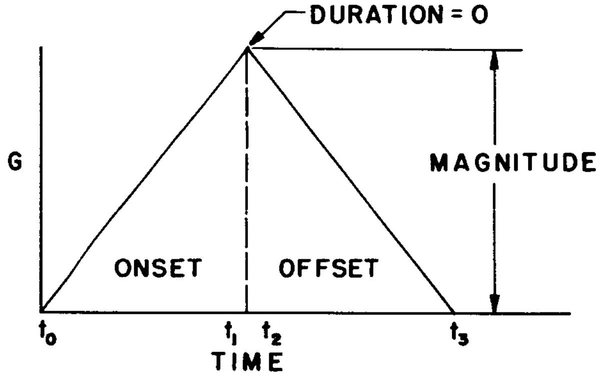


Figure 7. Example of a triangular pulse in which assessment of tolerability by comparison to the HIAD graph is not possible, since a plateau "duration" of .023 seconds or greater does not exist.

This statement is not intended to discredit the great work previously accomplished by Stapp(6), Beeding(7), and others, who purposely exposed themselves to dangerous thresholds of acceleration almost on points previously calculated by Lombard(8) to be intolerable. This work was accomplished at great risk of life and marked a milestone in our progress, since we were in a void at that time. We must hasten to state also that the test devices used by Von Gierke and Coerman, at ASD and the "Bopper" and "Daisy" Tracks at the Aero-Med Field Laboratory at Holloman Air Force Base, are unusually accurate and useful test machines, and yield useful information.

However, the aerospace industry is saying that experiments of the future should be undertaken only after the problem has been thoroughly analyzed by the dynamicist-mathematicians and, further, these people should be important members of the team, since the problem of impact acceleration, whether it be a human body or a piece of inanimate structure, is first a problem of dynamics and, second, a problem of physiological tolerance.

As Payne has so well stated(9), "What mathematics provides in this case is a logical measuring scale against which physiological phenomena can be correlated.

The sole function of the mathematician-dynamicist is to provide this measuring scale, convince research workers in the field that it is valid, and withdraw to the wing. He also has to provide instruction on occasion since medical facilities do not normally include mathematical dynamics in their curriculum and the subject cannot be absorbed by osmosis."

The first significant work in the field of human-body dynamics by the aerospace industry was by Kornhauser[10, 11] at General Electric Corporation. Since this is the subject of another paper at this symposium, a detailed discussion of his work will not be undertaken at this time. In summary, he has been able to demonstrate mathematically, and to a certain extent experimentally, that within the impact regime there is a band in which rate-of-onset and peak G are not of great significance, velocity change being the important parameter. He has also been able to show that we may expect a significant scatter in our test results using biological specimens. This work was largely company-funded and is of great importance to the bio-dynamics field.

Most of the other analytical research of recent years in industry has been motivated by the need to evaluate a specific man-machine system in lieu of usable, practical design allowables. One of these cases has been the Mercury Project. As in the B-58 Capsule, it became necessary for the McDonnell Aircraft Corporation to assess the landing capabilities of the Mercury Capsule in terms of human tolerance. A series of tests was performed using pigs to provide the initial evaluation of this system. The important thing, however, was not the actual pig experiments themselves, but the analytical work accomplished by Brock[12], who was able to create a mathematic model of the situation and obtain correlation between experimental and analytical data. His models were quite similar to those used by Kornhauser[11], Shapland[13], and others, and the field of bio-dynamics would suffer a great loss if we were deprived of further work by this dynamicist.

The writer has had the opportunity of working closely with two scientists during the past three years, from which work we may expect significant contributions in the impact-acceleration-tolerance field. These are, Peter Payne, Frost Engineering Development Corporation and Doctor Shapland, Stanley Aviation Corporation. Payne's work started with the problems associated with the ground landing system of the B-58 Capsule[14] and has contributed much to the analysis of this system. He has since contributed several worthy works pertaining to the dynamic analysis of impact-acceleration stress[15, 16, 17]. In addition, he has devised a computer for the specific purpose of analyzing the physiological effect of short-period acceleration on man which will be demonstrated at this symposium.

The idea of a computer for this purpose and the technology necessary to make it possible are not new. Only last month the writer watched a demonstration of a similar computer, designed and constructed by the Landair Corporation for the Aero-Med Field Laboratory at Holloman Air Force Base, with which it was possible to vary two parameters—frequency response and damping. Even this simple computer demonstrated the dynamic response of accelerometers, and in some cases the dynamic response of dummies used on the Daisy Track. In addition, a similar computer, being used at Stanley Aviation Corporation, was suggested by the work being accomplished under its NASA Contract. Even though the idea and technology necessary is not new, it is important that these developments are being carried forward to a working stage.

To best describe what computers may be able to do for us in terms of savings in research effort and, perhaps more important, savings in time during the design of a

specific weapon system, we should look at the methods presently used for measuring the response of a human body enclosed in a restraint system under the influence of impact acceleration. At the present time, we in the aerospace industry follow the same procedure as that followed by the services. We measure response of the human body by means of accelerometers which are strapped tightly around the chest and placed directly on the sternum. In addition, we sometimes place accelerometers on the hip, also strapped tightly to the body, when we are particularly interested in response of the body along the Z axis. We do this because we must rely solely on accelerometers to give us response information, and, since we are usually more interested in the response of the upper torso, the sternum appears to be the best place upon which to mount these accelerometers. However, we also know that accelerometers mounted in this manner do not always truly represent the accelerations experience by the torso, since they are affected by our inability to mount them firmly on the body as well as the flattening or flexture of the rib cage upon impact. Therefore, we are actually measuring the movement of the accelerometer itself and not necessarily the response of the torso.

Since tolerance basically involves the response of various organs in the body, and since this response could vary depending upon the confirmation of the subject and the type restraint environment in which he is placed, it would be necessary to measure experimentally the accelerations of each individual organ throughout all ranges enclosed in all types of restraint to assess tolerability in exact scientific terms. To arrive at allowables under all conditions by this method alone, therefore, would require more time than is available to us now in terms of national security. It should also be noted that the present published allowables now used for design purposes are not presented in terms of response, but as Eiband[3] and others[1, 2] have reported, they are presented in terms of a calculated input acceleration plateau. It can be proven mathematically and experimentally that with the same input acceleration plateau, the impact acceleration can become either tolerable or intolerable depending upon the response due to the restraint environment, natural frequency, damping, stiffness, and in some cases the bottoming out of cushioning materials(17, 18).

To summarize and emphasize the preceding statements again, ultimately, tolerability must be judged on the output or response of the body, and to obtain this information scientifically it would be necessary to instrument every vital organ in the body. If we had persisted in this approach in the field of aerodynamics, supersonic flight would no doubt still be a dream.

The general opinion of those having engaged in impact studies in the aerospace industry is that we will eventually be forced to calculate the output in the form of some type of physiological or response index, which would be equivalent to what Payne[17] calls the "idealized accelerometer." There is no question that this mythical index number, by whatever name we choose to call it, could be a very valid number, just as valid as the Reynolds number in aerodynamics or similar numbers obtained in other disciplines such as atomic physics.

As applied to the problem of response to impact acceleration we can find the essential parameters of any system in the laboratory, such as bottoming depth, equivalent stiffness, damping coefficients, natural frequency, etc. In addition, we have come far in our search for a dynamic model of the human body. This is the result of analytical research by Ruff[19], Kornhauser[4, 10, 22], Brock[12], Von Gierke, Coerman[23], Payne[14, 15, 16, 17], Shapland[13, 21], and others. This model is now undergoing refinement as a result of the reappraisal of the available human-tolerance-to-impact

data by Stanley Aviation Corporation under NASA Contract NASr-37. Therefore, once these parameters are known (and most can be obtained in a laboratory since they deal with mechanical properties primarily), the dynamicist and his computer can mathematically determine the true dynamic response under an infinite set of conditions equivalent to our "idealized accelerometer."

This capability is already at hand and gives us the following great advantages:

1. A well-defined program can now be planned with a definite objective in mind; thus, eliminating or more closely integrating the sometimes unilateral experiments now in progress by various agencies, universities, and industries.

2. Experiments can now be planned to correlate directly with this calculated index by placing biological specimens into a system with the exact mechanical properties as those inserted into the mathematical problem from which the response index was calculated.

It matters not to the index what the endpoint is, whether it be laceration of the liver or degradation of performance. The point is that it can be directly correlated with a method which would allow the engineer and the designer to design a man-machine system efficiently, since by the use of a simple computer he can vary mechanical parameters until he obtains an optimum system. When it becomes necessary (as in all engineering design) to compromise this system for the sake of other factors, he can assess this compromise by more exact methods.

Stanley Aviation Corporation's work, under NASA Contract NASr-37, is revealing significant information in obtaining a more realistic dynamic model of the human body without which a calculated output would be invalid. This work is also defining in more exact terms the existence of a band within the impact regime where rate-of-onset and peak G become less and less significant. It has also pointed the way toward the design of a more complex and sophisticated computer from which, by correlation even to some of the existing human tolerance to impact data obtained by other methods, it will be able to calculate human tolerance to impact under certain conditions. Doctor Shapland has described this work in more detail in another paper[21].

Experimental Impact Studies

Before proceeding to a description and discussion of actual experimental impact studies recently accomplished in aerospace industry, it should be pointed out that there are certain facilities in existence in industry which have not as yet been fully utilized.

One is the Columbus Division of North American Aviation, and consists of a vertical accelerator originally constructed under Navy funding for an evaluation of pilot performance when subjected to random gust and flight conditions in the A3J Air-plane, under the capable guidance of Robert Carter[25].

The other facility just recently constructed is the Life Science Research Laboratory at the Norair Division of Northrop, Hawthorne, California. Under the direction of Dr. Charles Lombard, this laboratory is exceptionally well planned and includes a miniature track and carriage similar to the "Bopper" track at Holloman Air Force Base, with which Doctor Lombard plans to do impact experiments of a basic nature. Among the experiments planned by Doctor Lombard are base-line studies to determine

why small animals, such as those used by Kornhauser, can withstand higher-impact accelerations in terms of velocity change than can man. Dr. Lombard is an early pioneer in acceleration research, with proven ability as attested by his acceleration work at University of Southern California in 1949[8] and later work in head-impact tolerance at Protection, Inc.[26], and the results he obtains will undoubtedly be useful.

Capsule systems have led to two recent programs in which biological specimens were subjected to impact when dropped from various heights and at various drift speeds at various attitudes when seated and restrained inside the capsule. One was on a modest scale due to the nature of the contract, and the other on a more elaborate scale due to the immediate operational requirements of that system.

The former was performed at the Los Angeles Division of North American Aviation under Hegenwald, who has contributed a worthy compilation of acceleration data[27] and who with Brockley, et al, obtained correlation between experimental, analytical, and medical data in the famous Smith ejection[28].

Hegenwald has performed a total of 17 capsule-learning tests using the XB-70 Escape Capsule; seven on decompressed granite, seven on reinforced concrete, and three in water. Of these 17 tests, four were conducted with human occupants (two on concrete and two in water) and six with anthropometric dummies. A weight capsule was used on the first seven and a production-type capsule was used on the last 10 tests. The velocity at impact was 28 feet per second vertically. On twelve tests, a horizontal drift condition equal to a 17-knot ground-wind condition was simulated. All of the tests were successful and the physiologic responses of the test subjects were satisfactory. Various test facilities were developed for the impact tests. Special rigged mobile cranes were utilized for vertical-drop and swing-drop tests. A 45-degree ramp was used to simulate capsule-landing on various ground surfaces.

The other, more sophisticated, program was performed to evaluate the landing characteristics of the B-58 Capsule by Stanley Aviation Corporation as a sub-contractor to the General Dynamics Corporation. Early in the design of the capsule-landing system, it became obvious that it would be impossible to evaluate its capabilities in terms of human tolerance by comparison with published human-tolerance allowables. A test program was instigated, therefore, to prove out the theory that extremely high peak G's and extremely high rates-of-onset could be sustained without injury provided the velocity change was below a certain level[29]. Twenty experiments were performed, including dropping human subjects on steel tables without attenuation, to prove this hypothesis.

This was followed by 132 drops using human subjects seated in a near-production B-58 Capsule, allowing the capsule to impact on various surfaces including sand, hard dirt, steel, and concrete, from various heights up to and including 28 FPS vertical velocity, and up to 20 knots ground-drift velocity on concrete. Since this program was not directly instigated by the Air Force, a minimal amount of medical surveillance was utilized. This consisted essentially of blood-pressure and pulse measurements before and after each test, and skeletal x-rays of each subject before and after use. No detectable injuries occurred, although peak G's up to 86.6 were recorded on the capsule. At that time, the capsule specification required a capability up to 20-knot drift at all attitudes, and a 28-FPS vertical velocity on concrete. This system was satisfactorily demonstrated with human subjects up to those conditions.

It became apparent, however, that landing on concrete using the B-58 Capsule attenuators was an easier condition to achieve than landing on soil. The Air Force, therefore, requested Stanley Aviation Corporation to embark upon another series of experiments under a separate contract, first determining the type of soil which would most likely impose the most hazard on the crewman, and then to perform experiments up to maximum conditions with biological specimens to assess tolerability[30]. A series of 25 experiments was made using human subjects at 25 FPS vertical velocity and up to 24 FPS horizontal velocity head forward, 25 FPS vertical velocity and up to 24 FPS horizontal velocity buttocks forward, 25 FPS vertical velocity and 35 FPS horizontal velocity right side forward. All subjects were males between the ages of 21 and 29, chosen at random from the unemployment bureau and subjected to the equivalent of a Type 3 Flying Status Physical Examination at Lowry Air Force Base prior to testing. Complete medical surveillance was exercised by attending medics from Lowry Air Force Base and privately employed physicians from Stanley Aviation Corporation.

The only injury sustained attributable to impact testing was a mild anterior compression fracture four mm in depth of thoracic vertebrae T3. The subject required no surgery and was released by the attending orthopedic surgeon after four weeks of observation.

Immediately following these tests, six tests were performed using American Black bears and Himalayan Black bears. These bears were subjected to 28 FPS vertical velocity and 36.5 FPS horizontal velocity in head-forward, buttocks-forward, and right-side-forward positions landing on hard dirt. Two Himalayan Black bears were then subjected to resultants, in terms of energy, representing 120 per cent of the preceding conditions in head-forward and buttocks-forward positions.

None of the bears received irreversible injuries, and no spinal fractures occurred on any test. One American Black bear was discovered upon autopsy to have a laceration of the liver, which was attributed partially to an overdose of anesthesia on that particular subject. This subject had received 52 cc's of Nembutal prior to dropping. One other bear died as the result of an impact, but the autopsy revealed that this bear was afflicted with an advanced case of hydrocephalus; no spinal damage had occurred. (Tables 1 through 8).

On October 2, 1961, a 99-pound female chimp inside a B-58 Capsule was fired from a sled at Hurricane Mesa at 600 knots, landing on a large boulder below the Mesa. The autopsy of this subject revealed no injuries. On November 10, 1961, an American Black bear inside a B-58 Capsule was fired from a sled at Hurricane Mesa at 693 knots, the capsule landing on soft dirt below the Mesa and rolling 360 degrees into a wash. The autopsy of this bear revealed no injuries. Autopsies were performed and documented by the Armed Forces Institute of Pathology, Washington, D. C.; The Army Pathology Section, Fitzsimons Hospital, Denver, Colorado; the Veterinarian Services Branch, Aero-Med Field Laboratory, Holloman Air Force Base; and Captain Neville Clarke, Aerospace Medical Laboratory, WADD, Wright-Patterson Air Force Base, Dayton, Ohio. All animal tests were accomplished with the support of the above services, including the Bio-dynamics Branch, Holloman Air Force Base.

Further human tests are now in progress at Stanley Aviation Corporation to evaluate means to prevent the previously reported spinal injury[31], and to evaluate back landing on a special reserve parachute worn by Air Force personnel scheduled to be ejected from the B-58 Airplane in the escape capsule at Edwards Air Force Base in December, 1961.

Conclusions and Recommendations

As a result of industry's considerable experience in trying to design to nebulous human tolerance-to-impact acceleration-stress allowables, and in view of this writer's observation that a significant loss in time and money occurs daily in industry as a result of this problem, the following recommendations are offered. These suggestions are offered in the hope that sufficient funding and effort is provided to carry them rapidly to a successful conclusion.

Planning

A special ad hoc panel should be formed, including not only medical scientists, dynamicists, and engineers within ARDC and the Navy, but also representatives from NASA, the Space Science Board, and industry. The primary function of this panel would be to plan future impact-acceleration research so that all experiments would be complementary and lead to achievement of common objectives. This implies a rigid standardization of experimental techniques, improved instrumentation, emphasis on the mathematical approach, and more use of computers. This panel should also instigate the preparation of an "Impact-Acceleration Handbook for Designers" to replace the present section in HIAD.

New Studies

Some of the most urgent studies needed at this time to relieve the bottleneck affecting efficient design are:

1. Further animal studies and experiments, including a comprehensive study of comparative anatomy and tissue strength, to obtain valid endpoint data without the use of human subjects. Present human-endpoint data in the impact regime is too sparse to be of statistical significance.

2. Studies to determine tolerance to complex accelerations, including abrupt pulses superimposed on sustained accelerations.

3. Intensive analytical studies to evolve usable mathematical techniques in order to facilitate design and reduce long, costly test programs.

TABLE 1

Z-Axis Drop Tests—No Attenuation

Run	Date	Subject Name	Subject Wt.	Drop Height In	Subject Spinal G's
1	3/3/60	Fink	193	7	18.7
2	3/3/60	Fink	193	7	23.0
3	3/3/60	Fink	193	11	21.0
4	3/3/60	Fink	193	11	16.0
5	3/3/60	Fink	193	13	21.6
6	3/3/60	Fink	193	13	29.7
7	3/3/60	Fink	193	16	23.6
8	3/3/60	Fink	193	16	22.0
9	3/3/60	Fink	193	16	26.2
10	3/3/60	Fink	193	17.5	26.5
11	3/3/60	Holcomb	139	7	14
12	3/3/60	Holcomb	139	13	25
13	3/3/60	Holcomb	139	14	25.2
14	3/3/60	Holcomb	139	16	25.2
15	3/3/60	Holcomb	139	14	31.7
16	3/3/60	Holcomb	139	16	38[illegible]

TABLE 2

Free Fall into Sand Box—Attitude 'A' (Heel)—No Attenuation

Run	Date	Subject			Drop Height	Subj.-Sp.	Subj.-Tr.	Cap.-Sp.	Cap.-Tr.
		Name	Wt.	Ht.	In.	MX.G	MX.G	MX.G	MX.G
1	4/19/60	M. Geyser	145	6'0"	22	9.4	9.8	6.3	7.8
2	4/19/60	M. Geyser	145	6'0"	27.5	11.8	14.0	11.4	10.5
3	4/19/60	O. Rotach	130	5'5"	33	-	-	-	-
4	4/19/60	O. Rotach	130	5'5"	33	12.0	15.9	6.2	10.6
5	4/19/60	L. Lemke	215	6'0"	39.5	17.6	16.3	15.1	14.0
6	4/20/60	J. Martin	175	5'11"	45	-	-	-	-
7	4/20/60	D. Boyer	160	6'2"	45	23.8	19.0	21.1	16.4
8	4/20/60	J. Martin	175	5'11"	50 3/4	15.0	21.1	14.9	17.1
9	4/20/60	D. Boyer	160	6'2"	59	24.7	20.5	12.8	16.1
10	4/20/60	J. Martin	175	5'11"	57	-	-	22.4	16.5
11	4/21/60	D. Boyer	160	6'2"	67	26.4	20.7	17.9	16.2
12	4/21/60	D. Boyer	160	6'2"	75	39.4	31.5	25.0	19.6
13	4/21/60	D. Boyer	160	6'2"	83	35.3	32.2	14.6	20.2
14	4/21/60	O. Rotach	130	5'5"	33	13.8	14.2	9.9	8.3
15	4/21/60	McMasters	150	5'8"	46	16.6	28.3	16.3	14.7
16	4/21/60	W. Truax	180	6'0"	51	15.0	21.0	15.4	18.1
17	4/21/60	T. Honor	168	5'8 1/2"	58 3/4	15.9	23.4	18.2	18.9

TABLE 3

Free Fall into Sand Box—Attitude 'B' (Back) Wedge-Attenuated

Run	Date	Subject Name	Subject Wt.	Subject Ht.	Drop Height In.	Subj.-Sp. MX.G	Subj.-Tr. MX.G	Cap.-Sp. MX.G	Cap.-Tr. MX.G
1	4/22/60	W. Truax	180	6'0"	34	5.0	13.3	Neg.	8.9
2	4/22/60	McMasters	150	5'8"	45	5.0	17.2	Neg.	12.2
3	4/22/60	T. Honor	168	5'8 1/2"	51	5.0	17.7	Neg.	12.6
4	4/22/60	D. Boyer	160	6'2"	62 1/4	6.0	28.2	4.2	15.0
5	4/22/60	W. Truax	180	6'0"	68	7.7	25.6	4.2	20.2
6	4/22/60	T. Honor	168	5'8 1/2"	72	8.4	25.6	Neg.	16.3
7	4/25/60	D. Boyer	160	6'2"	78	7.1	27.6	3.1	18.3
8	4/25/60	T. Honor	168	5'8 1/2"	84	7.2	30.2	4.1	26.4
9	4/25/60	D. Boyer	160	6'2"	90	12.4	31.3	4.1	29.8
10	4/25/60	T. Honor	168	5'8 1/2"	96	15.0	41.5	4.1	21.9
11	4/25/60	D. Boyer	160	6'2"	102	12.6	31.3	4.1	29.8
12	4/25/60	T. Honor	168	5'8 1/2"	108	11.4	41.5	4.1	32.7
13	4/26/60	T. Honor	168	5'8 1/2"	114	15.4	36.4	6.3	40.9
14	4/26/60	D. Boyer	160	6'2"	120	16.5	38.8	8.9	40.4
15	4/26/60	T. Honor	168	5'8 1/2"	126	21.2	53.2	6.8	35.8
16	4/26/60	D. Boyer	160	6'2"	132	16.2	51.7	9.5	50.4
17	4/26/60	T. Honor	168	5'8 1/2"	57	9.2	21.7	Neg.	18.2
18	4/26/60	T. Honor	168	5'8 1/2"	62 1/4	6.7	23.2	Neg.	18.3
19	4/26/60	R. Dodge	155	5'6"	96	12.8	42.6	Neg.	26.1
20	4/26/60	R. Dodge	155	5'6"	102	16.9	39.4	6.4	36.8

Neg.—Negligible

TABLE 4

Free Fall onto Hard Dirt—Attitude 'C' (Back—Head 9° Up)
Attenuation: Production-Type Ground Landing System

Run	Date	Subject			Drop Height	Subj.-Sp.	Subj.-Tr.	Cap.-Sp.	Cap.-Tr.
		Name	Wt.	Ht.	In.	MX.G	MX.G	MX.G	MX.G
1	6/17/60	T. Honor	168	5'8 1/2"	51	33.1	37.1	-	46.7
2	6/17/60	E. Walston	137	5'7"	62	33.9	53.7	-	41.8
3	6/17/60	N. Holsing	200	6'2"	72	36.7	34.4	-	45.4
4	6/17/60	T. Honor	168	5'8 1/2"	84	-	55.4	-	52.8
5	6/20/60	E. Walston	137	5'7"	90	-	60.4	-	55.0
6	6/20/60	N. Holsing	200	6'2"	90	-	53.0	-	63.2
7	6/20/60	T. Honor	168	5'8 1/2"	96	-	42.1	-	57.0
8	6/20/60	E. Walston	137	5'7"	102	-	50.6	-	55.8
9	6/20/60	T. Honor	168	5'8 1/2"	108	-	45.7	-	66.5
10	6/21/60	N. Holsing	200	6'2"	114	-	42.3	-	59.5
11	6/21/60	E. Walston	137	5'7"	120	35.8	56.6	-	59.5
12	6/21/60	T. Honor	168	5'8 1/2"	126	-	-	-	-
13	6/21/60	N. Holsing	200	6'2"	132	-	-	-	-

TABLE 5

Free Fall onto Steel Plate on Concrete--Attitude 'D' (Back-Head 9° Up)
Attenuation: Production-Type Ground Landing System

Run	Date	Subject			Height	Subj.-Sp.	Subj.-Tr.	Cap.-Sp.	Cap.-Tr.
		Name	Wt.	Ht.	In.	MX.G	MX.G	MX.G	MX.G
1	6/21/60	T. Honor	168	5'8 1/2"	60	-	42.1	-	64.3
2	6/21/60	E. Walston	137	5'7"	72	-	32.0	-	65.1
3	6/22/60	N. Holsing	200	6'2"	84 1/4"	12.8	36.1	-	66.5
4	6/22/60	T. Honor	168	5'8 1/2"	92	18.8	66.9	-	71.4
5	6/22/60	N. Holsing	200	6'2"	98	40.3	72.5	-	81.4
6	6/22/60	T. Honor	168	5'8 1/2"	108	32.8	47.8	-	68.8
7	6/23/60	T. Honor	168	5'8 1/2"	108	25.4	83.8	-	86.5
8	6/23/60	N. Holsing	200	6'2"	114	31.9	26.7	-	110.9
9	8/18/60	T. Honor	168	5'8 1/2"	84	-	38.3	-	-
10	8/18/60	T. Honor	168	5'8 1/2"	132	-	-	-	-
11	8/18/60	T. Honor	168	5'8 1/2"	144	-	63.5	-	-
12	8/25/60	T. Honor	168	5'8 1/2"	138	23.3	62.5	34.1	56.5

TABLE 6

Capsule Drops From Moving Truck

Drop Number	Date	Subject	Drop Height	Truck Speed	Accelerations: Subject Transverse	Subject Lateral	Subject Spinal	Capsule Transverse	Capsule Lateral	Capsule Spinal
1	8/26/60	Honor	9'9"	5	56.4	--	31.1	--	--	38.3
2	8/26/60	Honor	9'9"	8	27.8	--	14.9	--	--	17.8
3	9/14/60	Brown	9'9"	11	38.7	--	48.0	53.0	--	29.6
4	9/14/60	Honor	9'9"	14	64.7	--	68.5	68.0	--	38.8
5	9/15/60	Tower	9'9"	5	54.3	13.4	25.4	52.7	10.4	21.6
6	9/15/60	Tower	9'9"	11	--	--	--	--	--	--
7	9/16/60	Tower	9'9"	14	64.9	--	51.1	--	--	--
8	9/16/60	Tower	9'9"	17	61.9	--	57.1	--	--	--
9	9/28/60	Honor	9'9"	20	65.7	16.8	--	74.4	11.1	41.4
10	9/30/60	Tower	9'9"	24	65.3	10.2	--	56.5	15.1	29.1
11	10/11/60	Tower	12'	11	--	--	71.3	--	--	56.5
12	10/19/60	Honor	12'	17	67.8	--	59.3	--	--	--
13	10/20/60	Honor	12'	23	--	--	--	68.1	--	36.8
14	9/30/60	Honor	9'9"	11	65.1	53.8	40.5	70.5	27.6	32.2
15	10/4/60	Honor	9'9"	17	37.3	17.7	9.6	50.3	12.1	24.5
16	10/4/60	Nye	9'9"	23	63.2	7.4	10.4	46.7	11.5	37.1
17	10/7/60	Tower	9'9"	11	54.2	31.7	15.5	61.9	19.8	26.1
18	10/7/60	Nye	9'9"	14	55.9	22.5	13.8	66.8	18.5	28.0
19	10/10/60	Tower	9'9"	20	--	--	--	--	--	--
20	10/11/60	Nye	9'9"	23	61.2	--	13.2	--	--	--
21	10/21/60	Honor	12'	11	66.0	--	31.1	61.4	--	19.8

TABLE 7

Monorail Biological Drop-Test Results

Run#	Date	Subject	Orientation	Attitude	Drop Height	Horizontal Speed	Acceleration—Faired Peak G's: Subject Transverse	Subject Lateral	Subject Spinal	Capsule Transverse	Capsule Lateral	Capsule Spinal
5	3/2	Suzuki	Head Fwd	Normal	9'9"	8.0	No Instrumentation					
8	7/2	Evans	Feet Fwd	Normal	9'9"	6.0	Human Water Drops					
10	7/2	Suzuki	Side Fwd	Normal	9'9"	8.0						
30	6/2	Easter	Feet Fwd	Normal	9'9"	5.6		18.6	33.2	37.8	12.4	25.5
31	6/3	Larson	Feet Fwd	Normal	9'9"	8.0	51.6	28.0	24.8	34.6	11.8	19.3
32	6/3	Plam	Head Fwd	Normal	9'9"	5.9	97.0	12.6	-43.2	50.4	7.6	-17.7
33	6/5	Evans	Head Fwd	Normal	9'9"	8.9	91.0	14.6	-25.8	69.8	5.8	-15.5
35	6/5	Gillis	Side Fwd	Normal	9'9"	6.5	113.0	28.2	15.0	67.7	13.6	12.6
36	6/6	Bear	Feet Fwd	Normal	12'0"	22.6	59.7	38.6	50.6	57.6	19.7	32.4
37	6/6	Suzuki	Side Fwd	Normal	9'9"	8.9	93.8	48.4	33.7	55.4	21.1	19.8
38	6/7	Bear	Feet Fwd	Normal	12'0"	22.6	62.8	18.4	58.0	51.4	25.8	40.0
39	6/7	Maisel	Feet Fwd	Normal	9'9"	13.4	54.8	14.4	46.4	37.8	12.6	29.5
40	6/8	Easter	Head Fwd	Normal	9'9"	10.9	Instrumentation Failure					
41	6/8	Larson	Side Fwd	Normal	9'9"	6.5	81.4	65.0	50.4	40.0	19.5	16.6
42	6/9	Evans	Side Fwd	Normal	9'9"	15.7	49.8	50.6	13.8	40.0	25.6	15.8
43	6/9	Plam	Side Fwd	Normal	9'9"	23.1	41.8	62.4	16.1	34.3	31.5	13.7
44	6/9	Gillis	Side Fwd	Normal	9'9"	24.6	60.0	39.0	20.8	45.8	31.7	9.9
45	6/27	Bear	Head Fwd	Normal	12'0"	23.2	87.0	26.6	-60.4	28.8	14.1	-28.2
46	6/27	Maisel	Feet Fwd	Normal	9'9"	14.2	37.9	9.4	38.4	30.0	11.6	26.2
47	6/28	Nace	Feet Fwd	Normal	9'9"	10.0	51.8	10.8	64.2	38.4	9.1	21.6
48	6/28	Bear	Head Fwd	Normal	12'0"	23.0	62.0	8.6	-77.2	38.8	4.0	
49	6/29	Evans	Head Fwd	Normal	9'9"	14.0	56.7	11.6	-28.8	45.3	7.8	-19.5
50	6/29	Suzuki	Feet Fwd	Normal	9'9"	16.0	43.5	11.4	43.6	37.4	11.7	-28.8
51	7/6	Nace	Head Fwd	Normal	9'9"	15.6	126.5	10.0	-79.2	86.6	17.7	-28.8
52	7/6	Larson	Side Fwd	Normal	11'0"	9.3	103.8	39.2	27.4	53.0	23.2	30.8
53	7/7	Evans	Head Fwd	Normal	9'9"	16.0	70.8	12.6	-21.6	52.4	9.8	3/4 26.0

TABLE 8

Monorail Biological Drop-Test Results

							Acceleration—Faired Peak G's					
					Drop	Horizontal	Subject			Capsule		
Run#	Date	Subject	Orientation	Attitude	Height	Speed	Transverse	Lateral	Spinal	Transverse	Lateral	Spinal
58	7/11	Evans	Side Fwd	Normal	11'0"	10.7	67.8	43.3	39.7	63.3	27.5	20.0
59	7/12	Nace	Side Fwd	Normal	11'0"	14.5	98.5	56.6	43.6	78.7	31.0	24.2
60	7/12	Larson	Side Fwd	Normal	11'0"	16.0	45.2	41.8	21.8	47.7	42.4	21.2
61	8/29	Bear	Side Fwd	Normal	12'0"	23.0	43.3	48.8	27.6	51.7	23.7	16.8
62	8/31	Bear	Feet Fwd	Normal	14'0"	24.8	58.1	33.2	74.6	37.4	8.5	32.4
63	9/12	Bear	Head Fwd	Normal	14'0"	24.8	46.5	34.2	-36.2	46.8	15.0	-27.9
64	9/14	Bear	Feet Fwd	Normal	14'0"	24.7	75.7	30.3	64.1	47.8	10.7	33.8

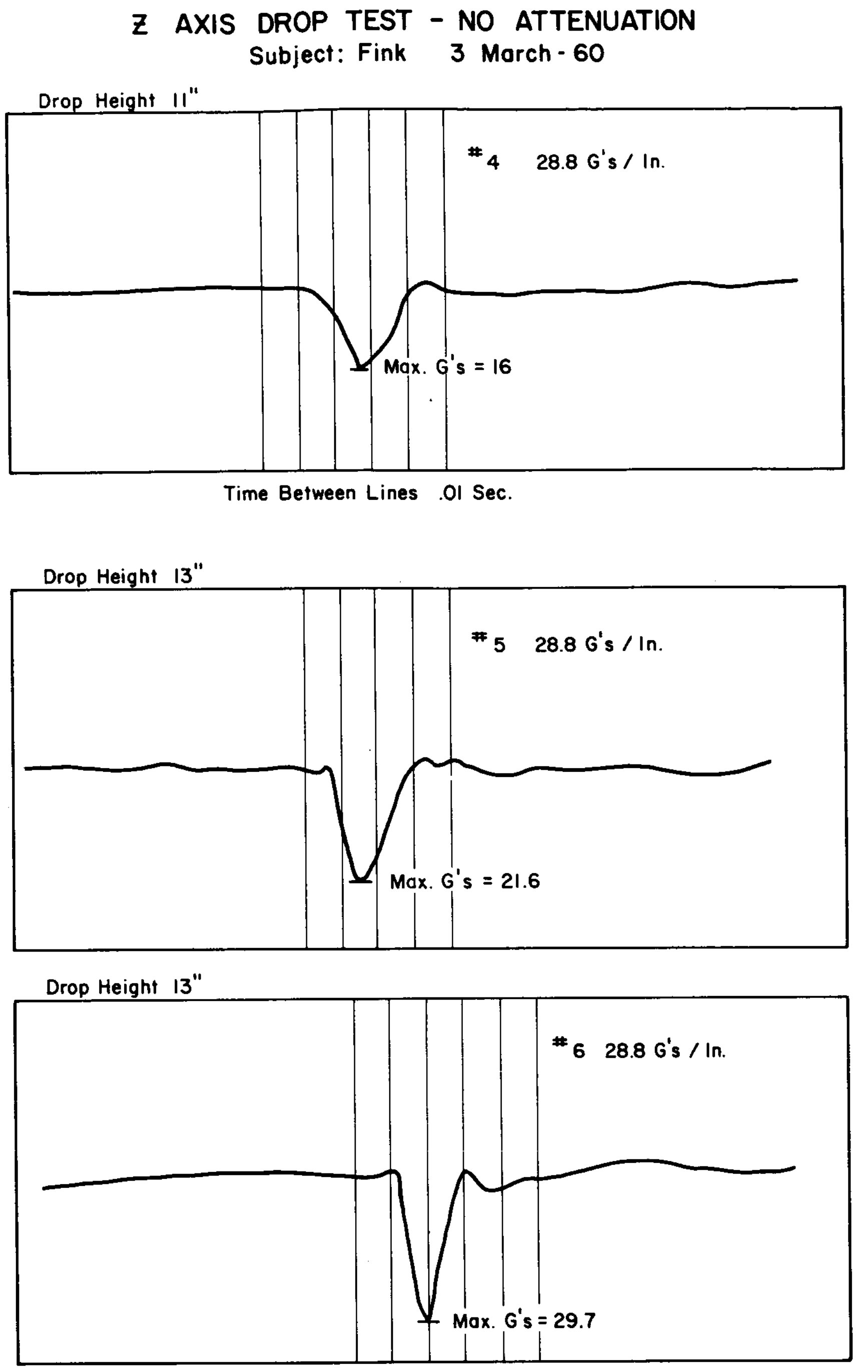

Z AXIS DROP TEST - NO ATTENUATION
Subject: Fink 3 March - 60
Drop Height 11"
#4 28.8 G's / In.
Max. G's = 16
Time Between Lines .01 Sec.
Drop Height 13"
#5 28.8 G's / In.
Max. G's = 21.6
Drop Height 13"
#6 28.8 G's / In.
Max. G's = 29.7

Figure 8

Z AXIS DROP TEST - NO ATTENUATION
Subject: Holcomb 3 March-60

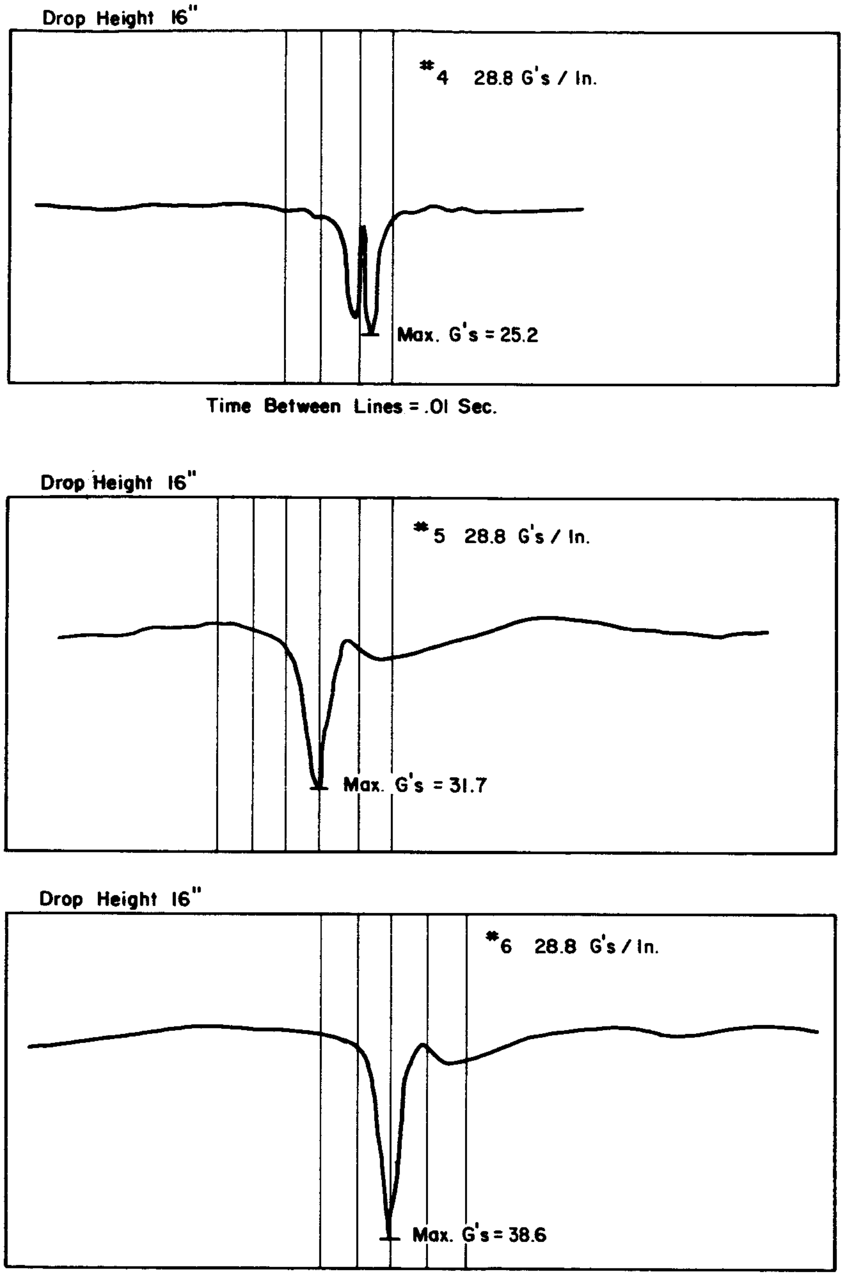


Figure 9

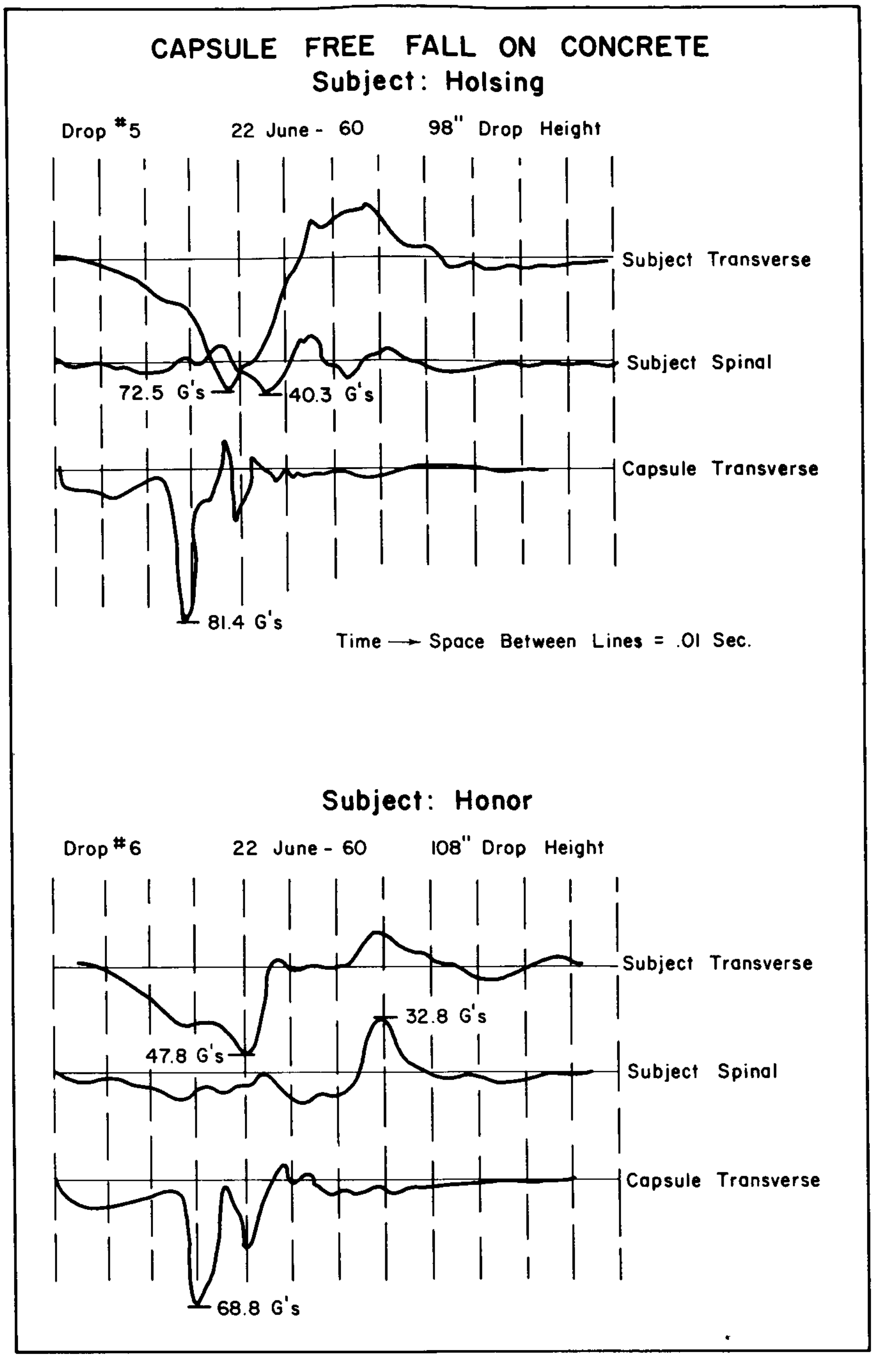

CAPSULE FREE FALL ON CONCRETE
Subject: Holsing
Drop #5
22 June - 60
98" Drop Height
Subject Transverse
Subject Spinal
72.5 G's
40.3 G's
Capsule Transverse
81.4 G's
Time → Space Between Lines = .01 Sec.
Subject: Honor
Drop #6
22 June - 60
108" Drop Height
Subject Transverse
32.8 G's
47.8 G's
Subject Spinal
Capsule Transverse
68.8 G's

Figure 10

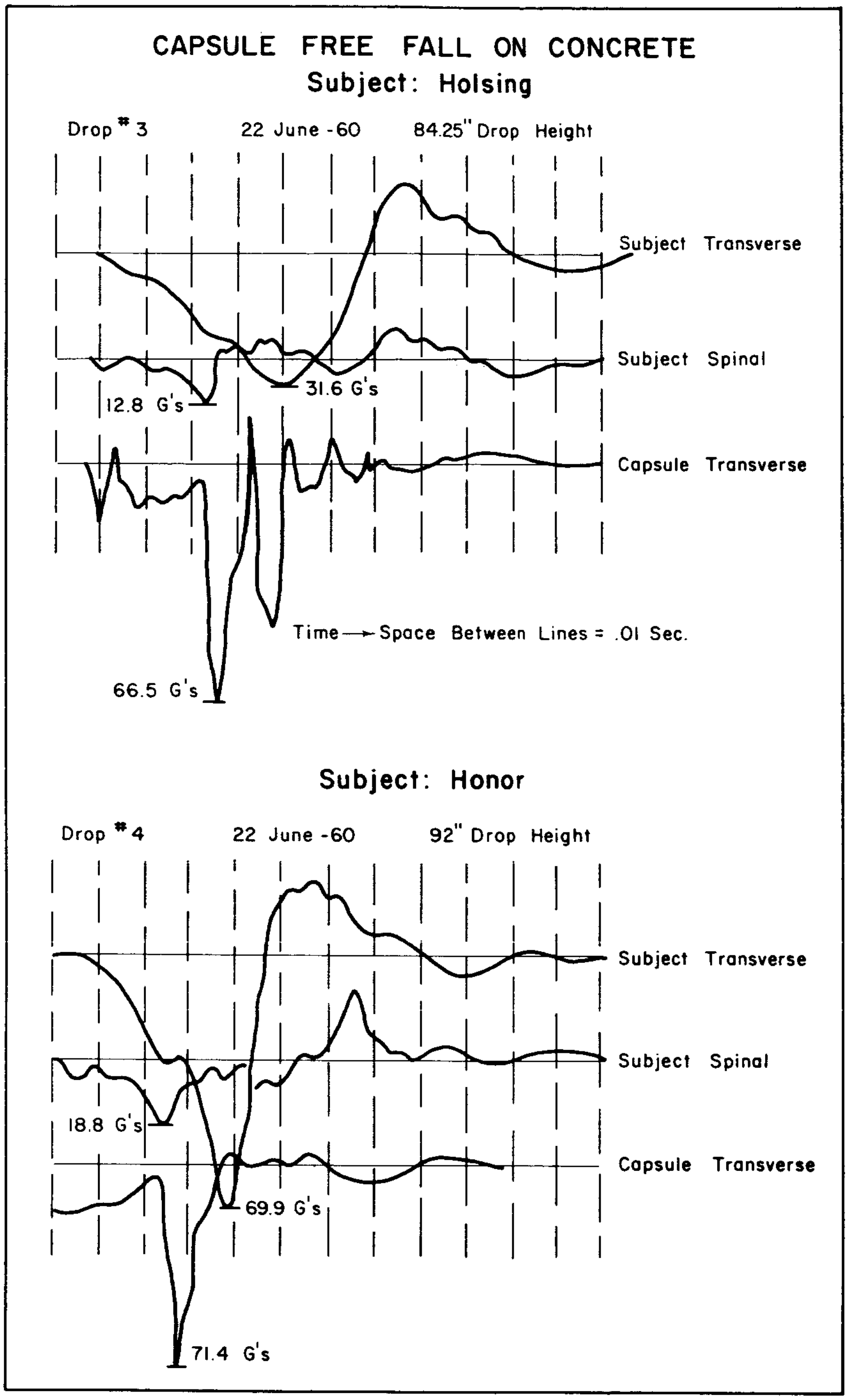
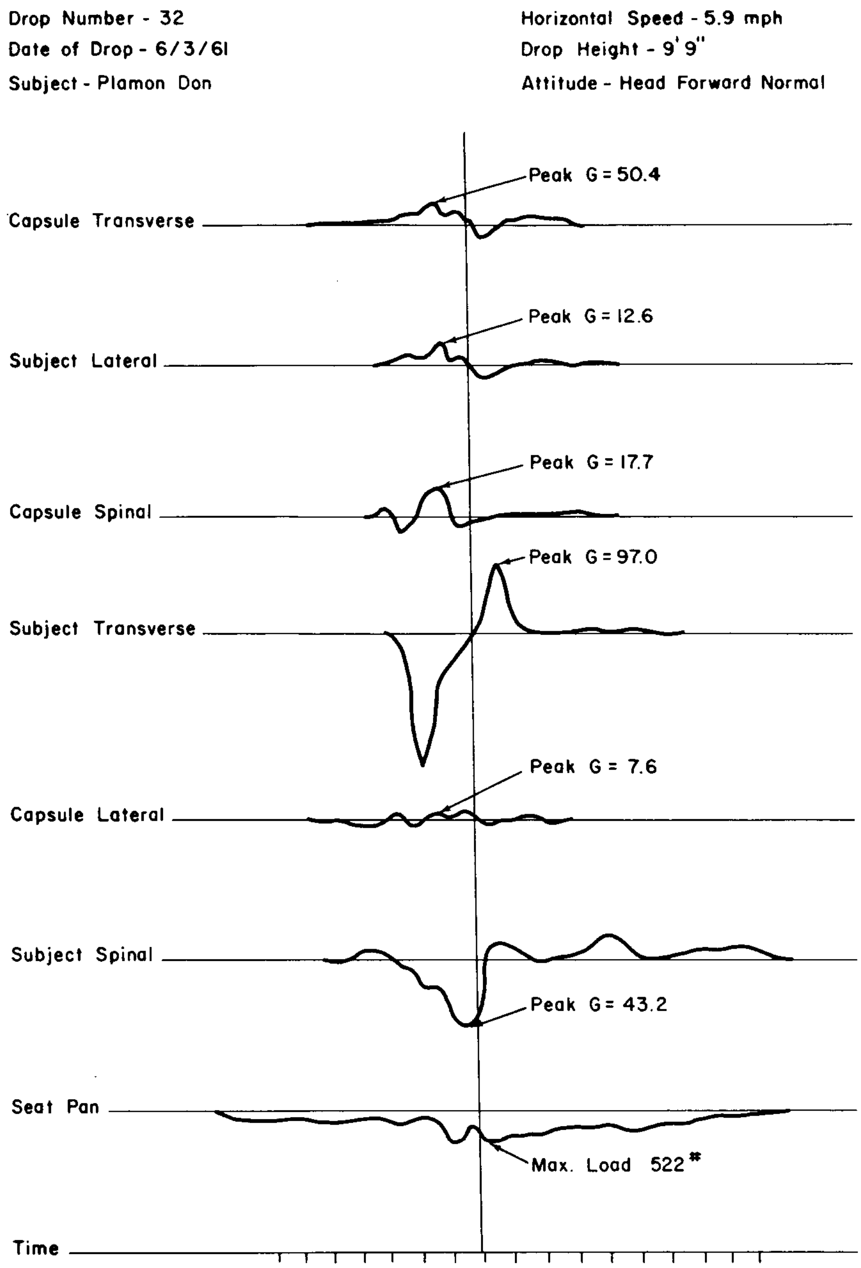

CAPSULE FREE FALL ON CONCRETE
Subject: Holsing
Drop # 3
22 June -60
84.25" Drop Height
Subject Transverse
Subject Spinal
12.8 G's
31.6 G's
Capsule Transverse
Time → Space Between Lines = .01 Sec.
66.5 G's
Subject: Honor
Drop # 4
22 June -60
92" Drop Height
Subject Transverse
Subject Spinal
18.8 G's
Capsule Transverse
69.9 G's
71.4 G's

Figure 11



Drop Number - 32
Date of Drop - 6/3/61
Subject - Plamon Don
Horizontal Speed - 5.9 mph
Drop Height - 9'9"
Attitude - Head Forward Normal
Peak G = 50.4
Capsule Transverse
Peak G = 12.6
Subject Lateral
Peak G = 17.7
Capsule Spinal
Peak G = 97.0
Subject Transverse
Peak G = 7.6
Capsule Lateral
Subject Spinal
Peak G = 43.2
Seat Pan
Max. Load 522 #
Time

Figure 12

Drop Number - 33
Date of Drop - 6/5/61
Subject - Evans

Horizontal Speed - 8.9 mph
Drop Height - 9' 9"
Attitude - Head Forward Normal

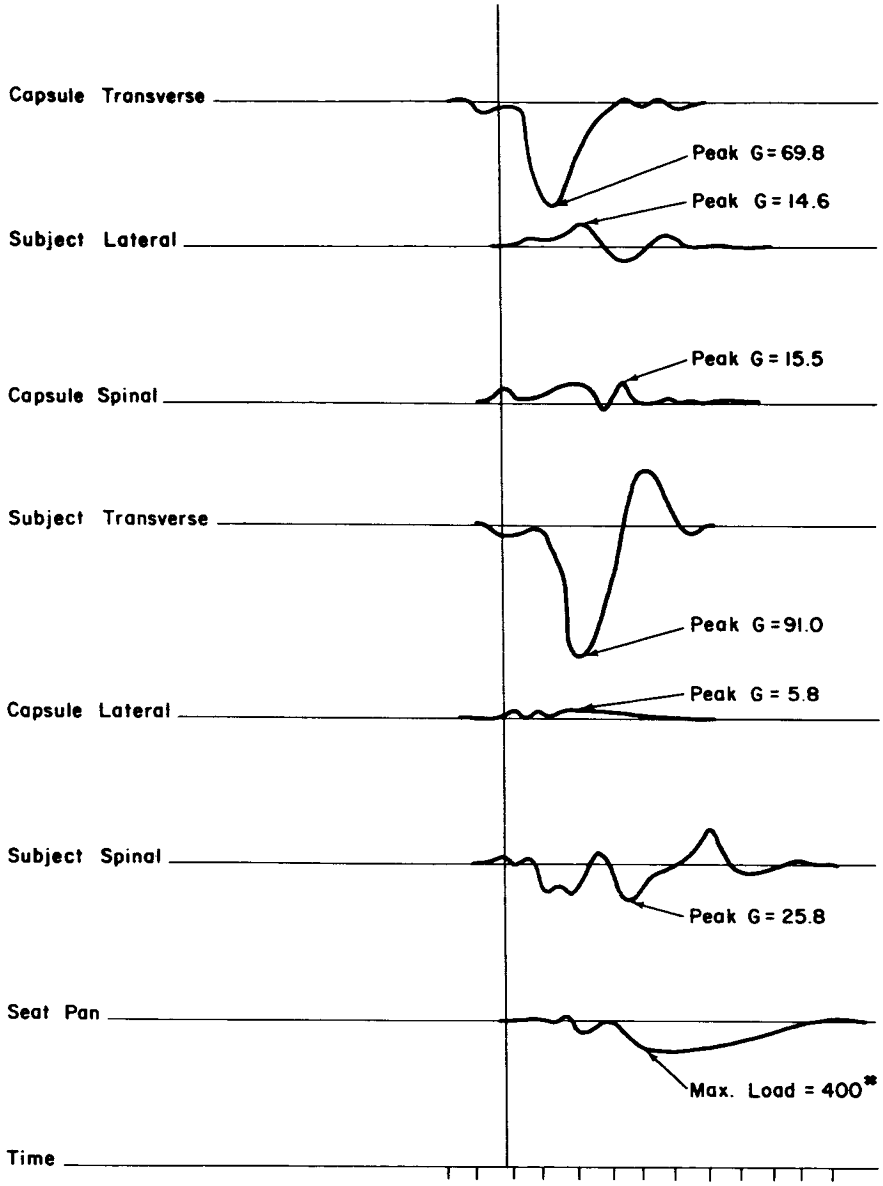


Figure 13

Drop Number - 35
Date of Drop - 6/5/61
Subject - Gillis

Horizontal Speed - 6.5 mph
Drop Height - 9' 9"
Attitude - Side Forward Normal

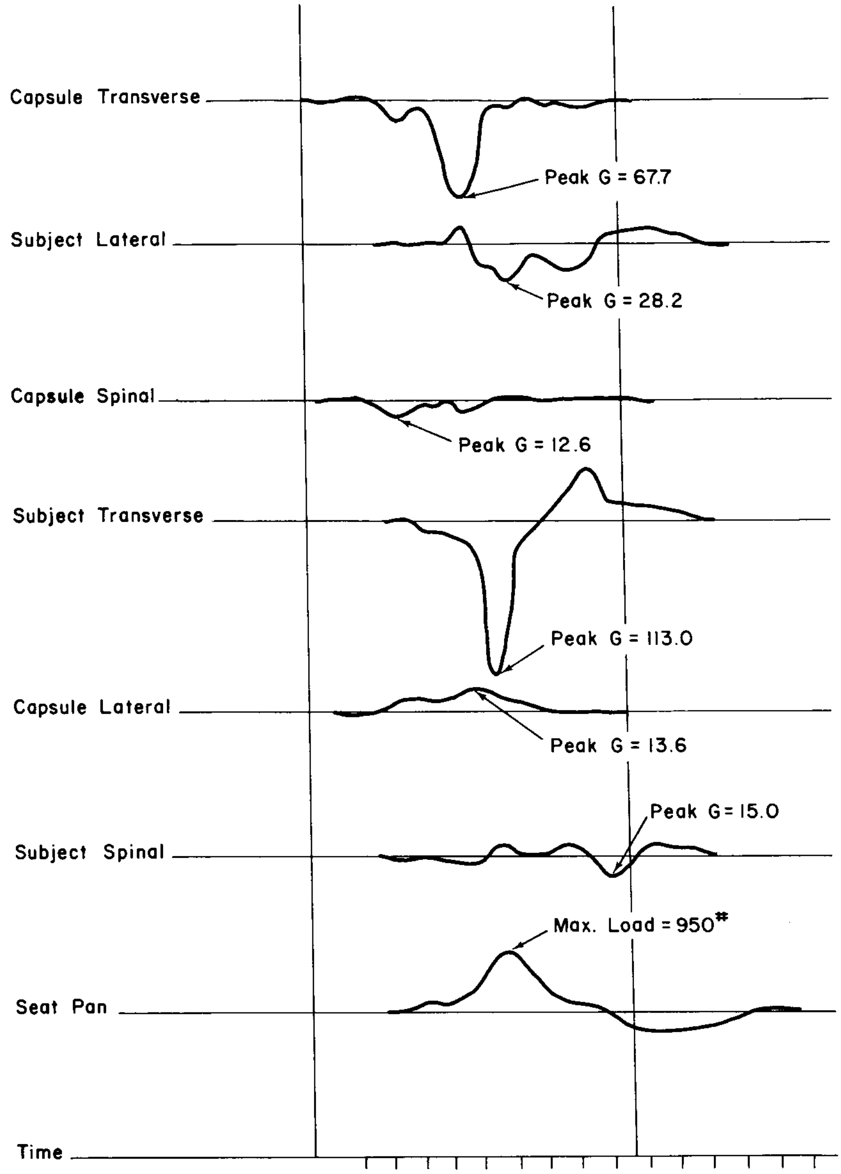


Figure 14

Drop Number - 59
Date of Drop - 7/12/61
Subject - Nace

Horizontal Speed - 14.5 mph
Drop Height - 11'
Attitude - Side Forward Normal

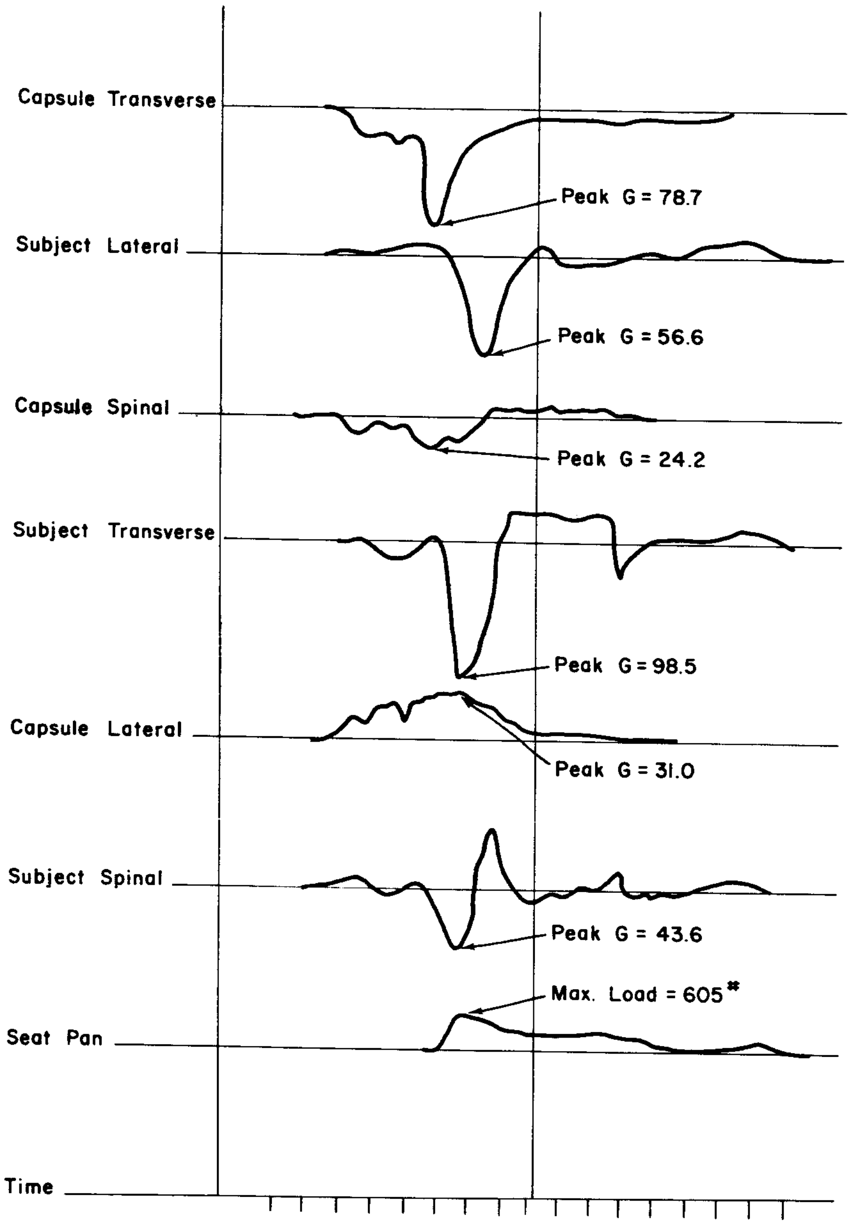


Figure 15

Drop Number - 61
Date of Drop - 8/29/61
Subject - Bear

Horizontal Speed - 23 mph
Drop Height - 12'
Attitude - Side Forward Normal

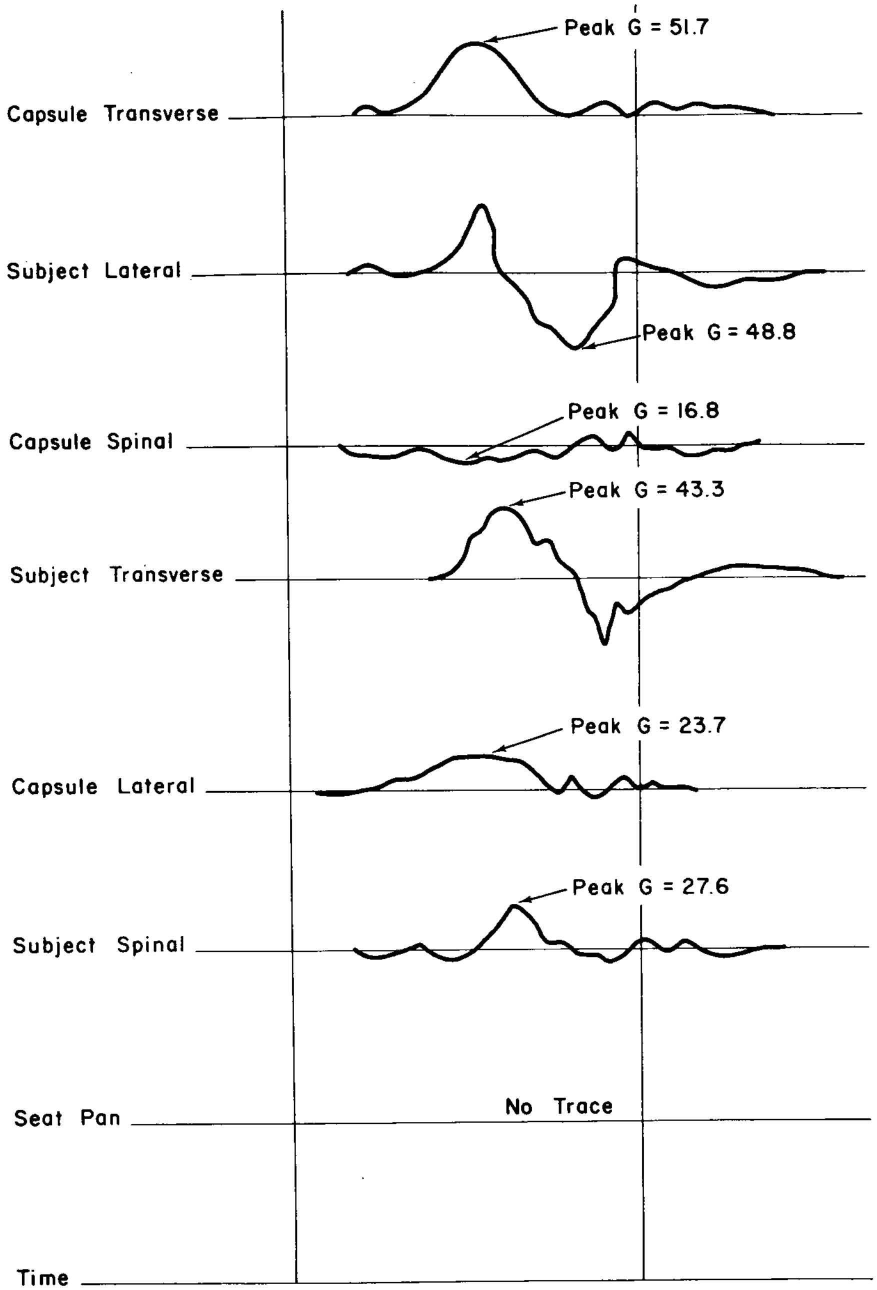


Figure 16

Drop Number - 64
Date of Drop - 9/14/61
Subject - Bear

Horizontal Speed - 24.7 mph
Drop Height - 14'
Attitude - Feet Forward Normal

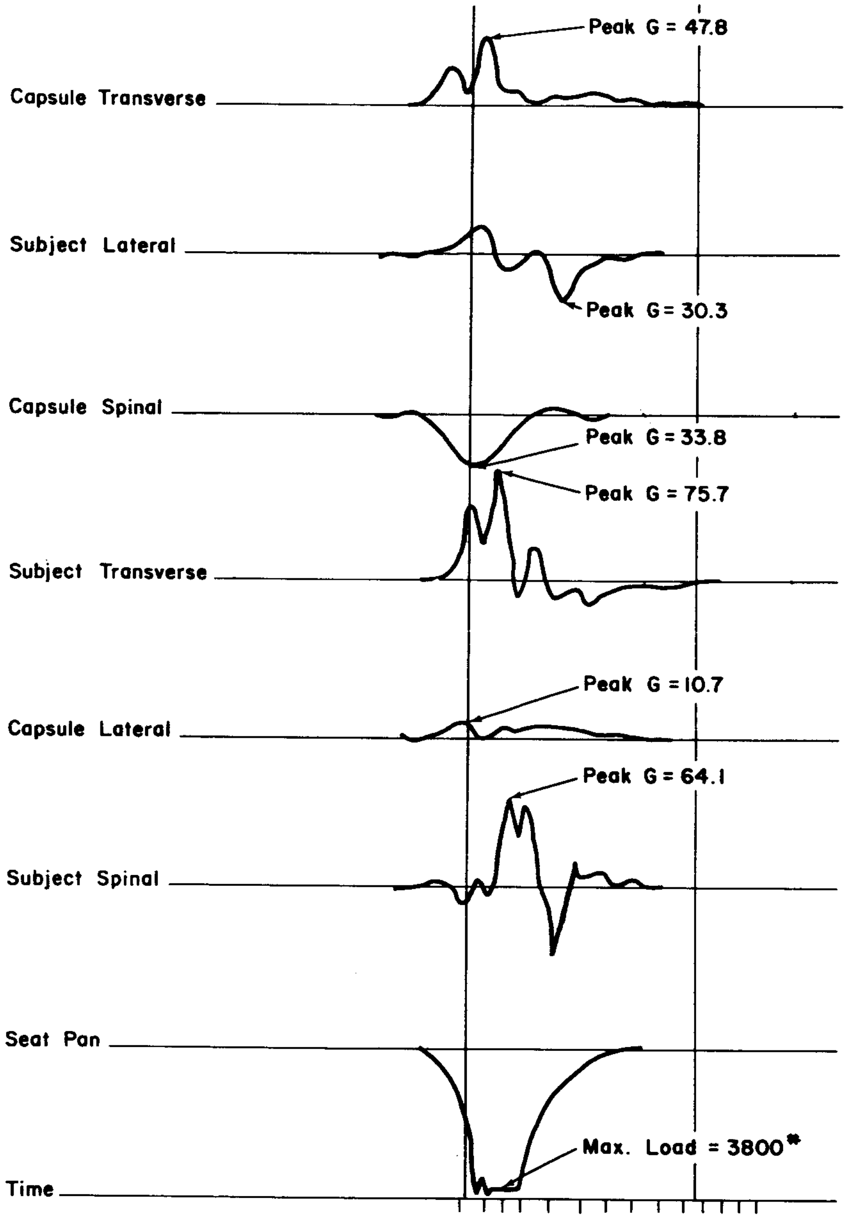


Figure 17

Figure 18. Chimpanzee test subject for HSRS sled run.

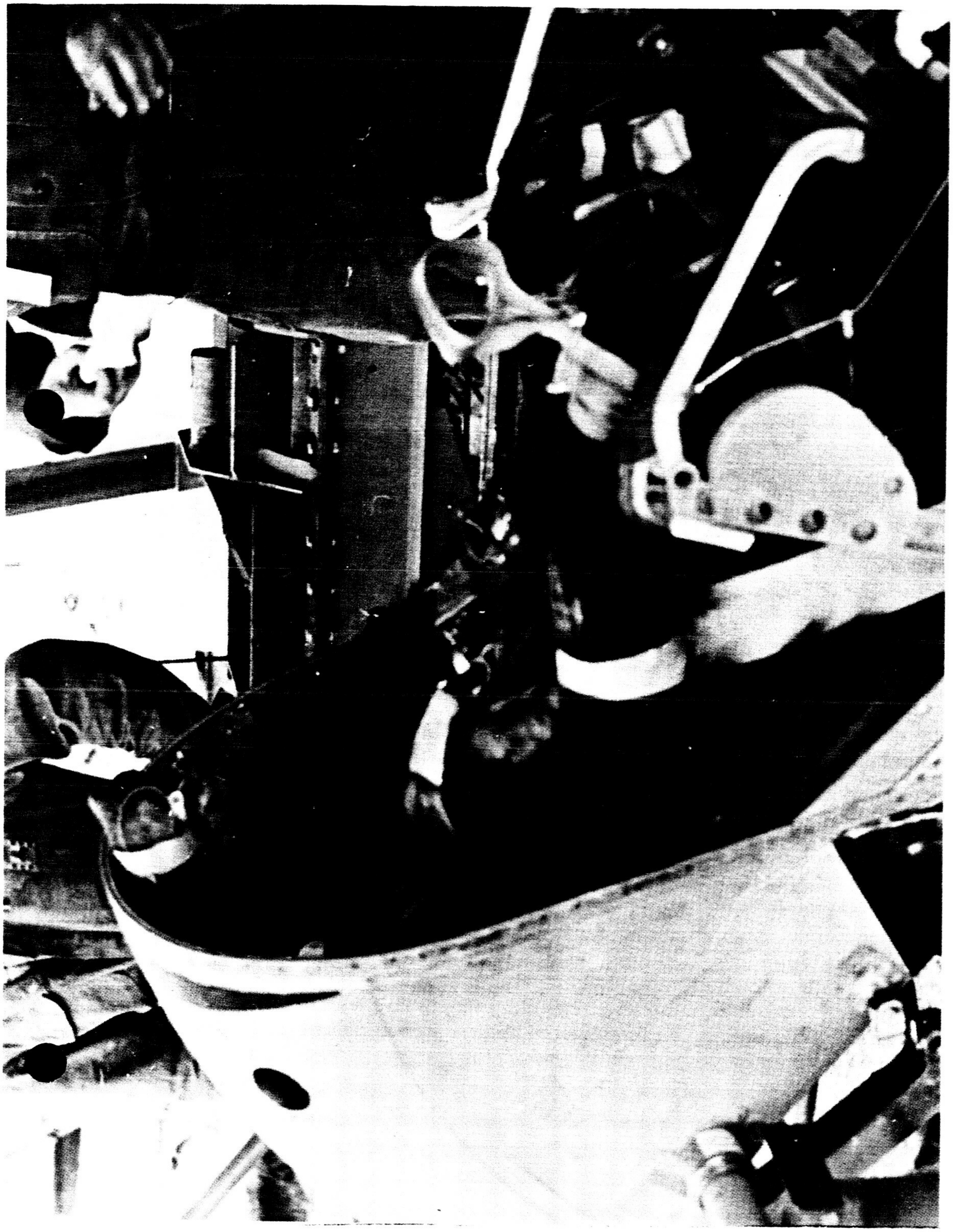


Figure 19. Black bear test subject for monorail drop test.

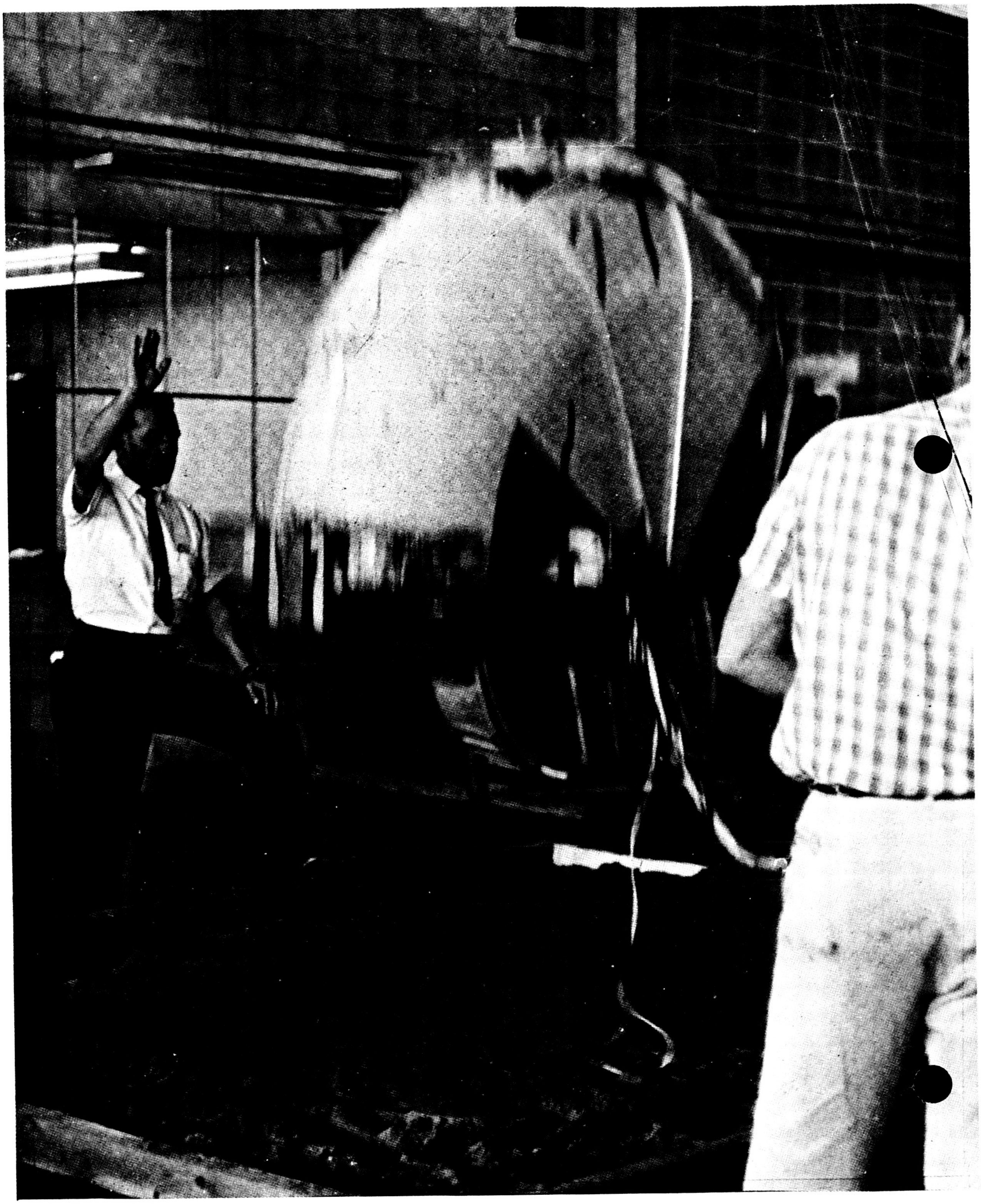


Figure 20. Capsule static drop test.

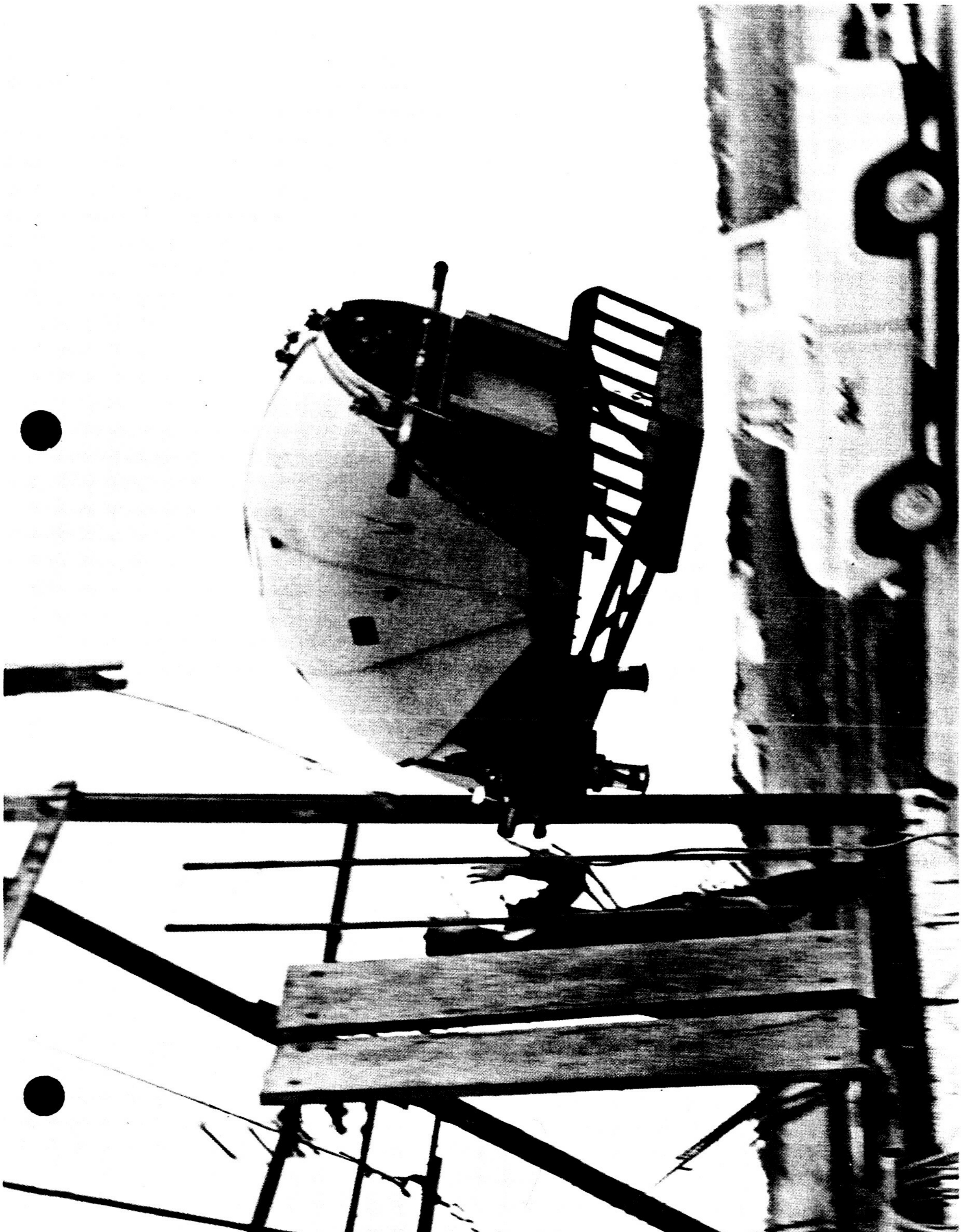


Figure 21. Capsule drop test from moving truck.

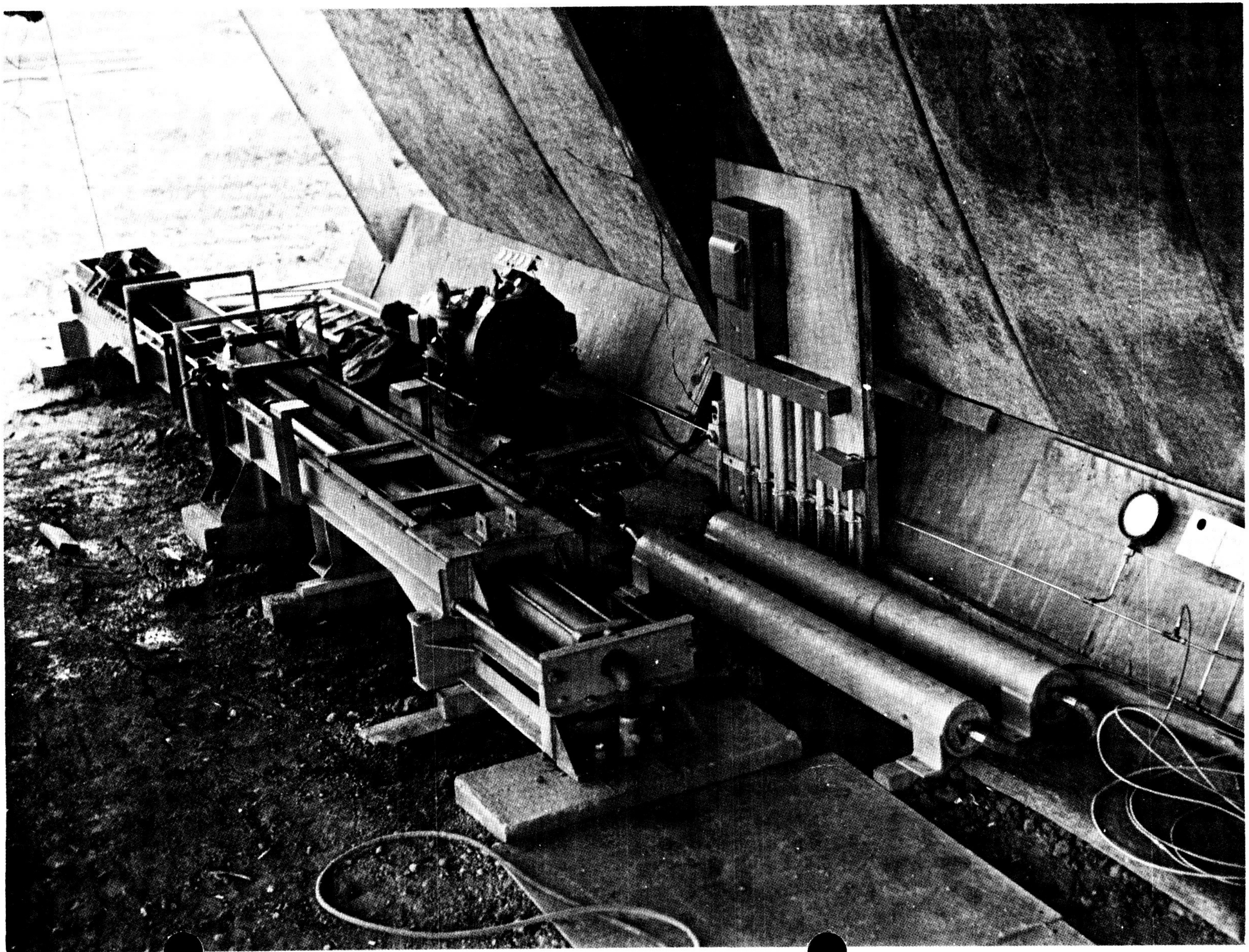


Figure 22. Monorail test facility catapult mechanism.

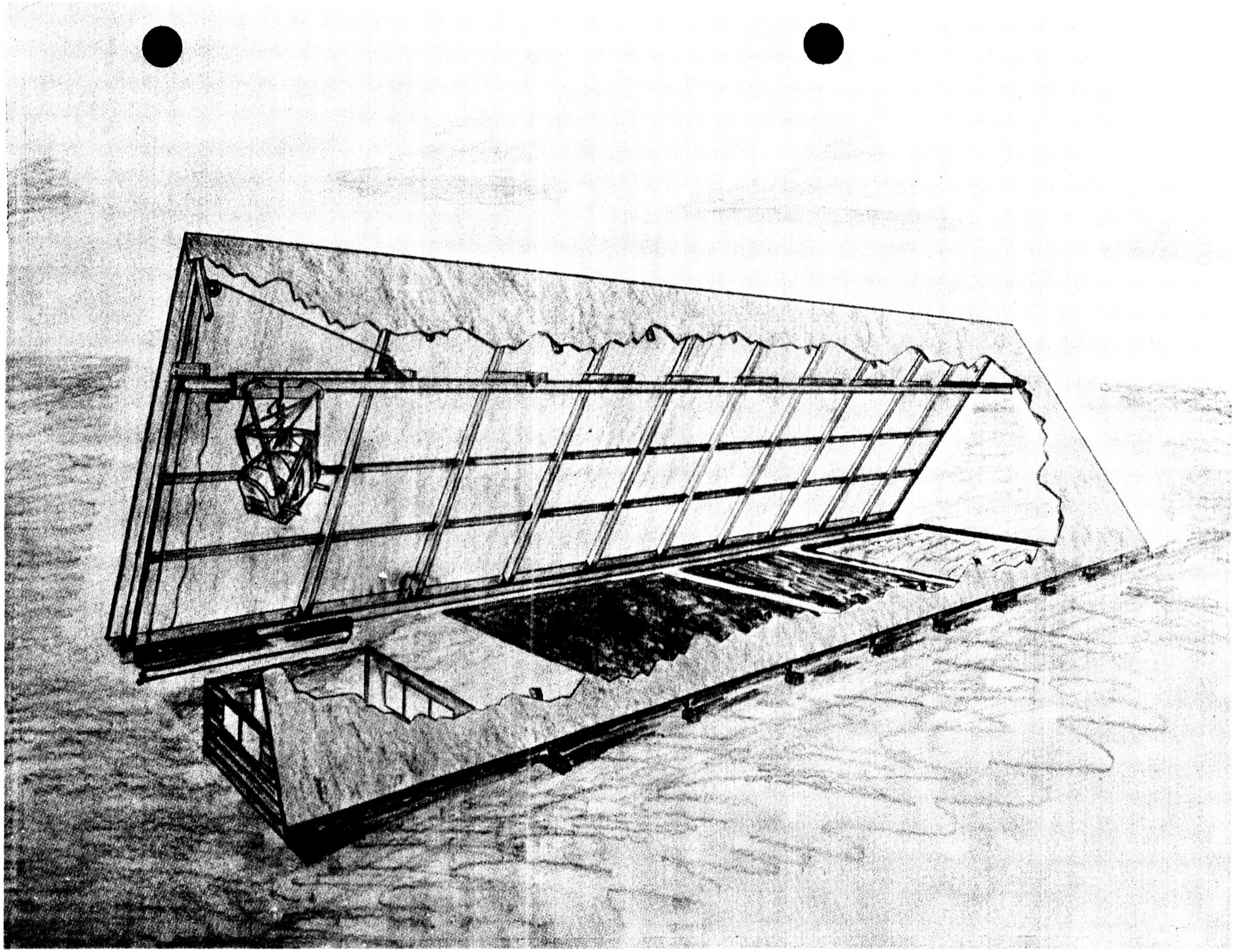


Figure 23. Artist's conception of monorail test facility.

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12856

BIOMECHANICS OF IMPACT INJURY

Henning E. von Gierke
Aerospace Medical Laboratory
Wright-Patterson Air Force Base, Ohio

This paper reviews the dynamics of the human body under impact-acceleration stress and discusses the body's response to impact in relation to its response to low-frequency, steady-state vibration. For excitation with steady-state vibration and with abrupt acceleration patterns containing frequencies below approximately 50 cps, the response can be approximated by analyzing the body as a linear, passive, lumped-parameter system. Recent studies contributed to a further refinement of such a mechanical model and to a quantitative determination of its parameters. For impact as well as for steady-state excitation of this type of system one can derive sets of curves characterizing, as a function of excitation frequency or pulse duration, equal mechanical stress or relative displacement for each body part or subsystem identified in the model. The various physiological and pathological phenomena limiting human tolerance to vibration and impact loads are caused ultimately by a combination of such mechanical stresses. However, the problem remains of determining at what levels mechanical stress or physiological tolerance limits are reached, and whether our present model is fine enough to identify those tissue areas most sensitive to mechanical insult.

The foregoing considerations are applied to the case of a sitting subject exposed to longitudinal loads. The simplified model for the subject is characterized in this case by an abdominal, lumbo-sacral, and head resonance in order of increasing resonance frequency. Tolerance curves to steady-state vibrations are composites of the tolerance curve for each individual subsystem, such as abdomen or head. Tolerance levels are generally minimum at the resonance frequencies of these subsystems. Similarly the overall impact-tolerance curve is a composite of the tolerance curve for each subsystem. For each of these there are impact-duration ranges where the system's response depends primarily on the pulse length, or the pulse shape independent of the pulse length, or the pulse integral (impulse). Therefore, theoretical analysis of the response of the complex human system to impact loads shows clearly that a complete description of the force-time function of the impact load is necessary to define response or tolerance uniquely. Only in very limited impact-duration ranges can single parameters such as peak acceleration, impulse, or rate-of-onset be expected to correlate as being primarily responsible for the response. Such considerations are roughly in agreement with presently available data on vibration and impact tolerance, and allow logical interpretation of available results.

Further refinement of existing models and study of other load directions in addition to the longitudinal condition are necessary. Knowledge of the dynamic properties of the human body, of the bodies of various animals, and of dummies is required to draw valid conclusions from animal or dummy experiments for extrapolation to human subjects and for analyzing protective systems and restraints as part of the overall system. However, in spite of the success of lumped-parameter analysis in interpreting the gross phenomena of biological-tolerance data, it must be kept in mind that this

approximation is valid only at the very low frequencies. Shear and compression waves become more and more pronounced as the excitation frequency is increased, and can be the cause of tissue damage at locations not predicted by lumped-parameter analysis. Theoretical analysis in its present state is therefore extremely helpful for evaluating the relative injury potential of various force functions, in interpreting biological effects, in guiding future experimentation, and in developing and understanding protective measures. However, actual human tolerance experiments will remain indispensable as the final decisive test for a long time to come. They can be safely replaced by animal experiments only when proper and more precise dynamic and physiological scaling laws have been established.

The following conclusions and recommendations are made:

1. Steady-state vibration studies are very helpful for interpreting impact-tolerance data. The mechanical models for the human body derived from such studies are useful as a basis for theoretical analysis and prediction of impact response.

2. Theoretical analysis of the response of the complex human system to impact loads shows clearly that a complete description of the force-time function of the impact load is necessary to define response or tolerance uniquely. Only in very limited impact-duration ranges can a single parameter such as peak acceleration, impulse, or rate-of-onset be considered primary to the response. Future experimentation, protection efforts, and instrumentation should be guided by the entire wealth of theoretical knowledge available to date.

3. Refinement of the models for the human body and study of the dynamic properties of animals, dummies, and various restraint systems is necessary to increase the quantitative value of mathematical predictions. Present limits of this approach are determined not by the lack of theoretical tools or the complexity of the system, both of which can be overcome by computer techniques, but by the lack of reliable and refined measurements.

4. Impact research should be conducted as an integrated part of an overall biodynamics program studying all aspects of the interaction of mechanical energy with the human body. Only such a broad approach will provide the capability to deal with the full complexity of present and future problems, not only empirically but also analytically.

12557

JOLT EFFECTS OF IMPACT ON MAN

Col. John P. Stapp
Aerospace Medical Division (AFSC)
Brooks AF Base, San Antonio, Texas

In aircraft passenger safety, human tolerance to impact forces has been underestimated or disregarded in the design of seats, their attachments, their optimum positioning in the aircraft, and passenger restraints. In connection with escape from military aircraft, there is a tendency to over-reach human tolerance with the impact loads designed into ejection devices. Requirements for human tolerance criteria to avoid hazard from either of these extremes led to empirical experiments on exposure of human volunteers to controlled increments of impact forces. Vertical-ejection simulators and occasional experimental ejections from aircraft provided data for escape-system design criteria; and various types of linear decelerators, such as rocket sleds, catapult sleds, snubbed swing devices, and drop-test towers have been used to obtain data for safety design to protect seated occupants of aircraft.

Human exposures to known impact hazards include: (1) ground-vehicle-crash impact, (2) aircraft-crash forces, (3) abrupt vertical accelerations of ejection devices to escape from aircraft, (4) windblast impact during supersonic ejection from aircraft, (5) high-velocity parachute-opening shock, (6) landing impacts on ground or water with or without seat or capsule, (7) high localized-impact forces due to collision with solid objects, such as occur in a fall with the head striking pavement, (8) blast forces encounter in aerial or underwater explosions, and (9) penetration impacts of high-velocity projectiles, such as bullets or shrapnel.

In all these cases, the effects result from an interaction of linear, quadratic, and cubic factors of motion with inertial factors complicated by resonant modes, elastic deformation, viscous damping, and structural complexity. In terms of the fundamental concepts of time and distance, the motion factors are defined as time-distance derivatives:

Where l = distance, and t = time

l/t = velocity
l/t^2 = acceleration, or velocity change with time in either rate or direction
l/t^3 = jolt, or rate of change of acceleration.

Mass set into motion has corresponding inertial derivatives:

Where m = mass, l = distance, and t = time

ml/t = momentum;	ml^2/t = action
ml/t^2 = force;	ml^2/t^2 = work, energy
ml/t^3 = onset;	ml^2/t^3 = power.

Linear, quadratic, or cubic rates of displacement acting on a mass produce equal and opposite reactions in the corresponding inertial functions of momentum, force, and onset, which acting through distance are measured as action, energy, and power. A body having elastic deformation will have two components of reaction to being set in motion: (1) a constant quantity corresponding to the static force required to displace the mass through a given distance, known as the forced response, and (2) a variable quantity relating to force applied to the elastic deformation, known as the free response. The free response is a function of the alternate deformation and restoration of the elastic property of the body, which has a rate known as the natural frequency, determined by the inertia of the mass reacting with the elastic oscillation of the body. The natural frequency of this free response presents a phase relationship to forces setting it in motion; the rate of application, magnitude, and duration of force disturbing the elastic property will affect the amplitude of free response. The sum of the forced response and the free response is the dynamic response; for a given rate and duration of applying a force or load, there will be a maximum dynamic response, which is a function of the natural frequency and is called the dynamic load factor.

In the case of instantaneous loading for a duration slightly exceeding one-half the natural frequency period, free response becomes maximum and equal to forced response, and their sum is a dynamic load factor of 2.0. For instantaneous loading lasting less than half the natural frequency period, the dynamic load factor is the ratio of load duration to natural period, diminishing from 2.0 toward zero as a limit.

Where loading is not instantaneous, but a triangular pulse, the maximum dynamic load factor occurs at slightly less than one-fourth the duration of the natural period and has a value of 1.52. As duration increases beyond a fourth of the natural period, the dynamic load factor becomes progressively smaller.

For a load applied as the first half of a sine wave, a duration matching the natural frequency of the body has the largest dynamic load factor, for a value of 1.75.

A blast pulse consists of a positive overpressure followed by a negative pressure. Where the body has a natural period less than 4/3 the duration of blast pulse, the maximum response factor occurs during the overpressure, and if the natural period is greater than 4/3 the positive blast pulse, the maximum response occurs during the negative pressure following the overpressure.

All of these cases are ideal for a single linear response, corresponding to a system with one degree of freedom. In systems of many degrees of freedom, dynamic load factors can vary with location in the structure and may considerably exceed the value of 2.0.

Damping is a resistance to deformation and restoration that diminishes oscillation at the natural frequency, and limits free response.

Actual structures approximate a single system with one degree of freedom for the fundamental frequency if the duration of impact is a tenth or more of the fundamental natural period, loading of structure is fairly uniform, and higher frequencies are well separated from fundamental mode. The boundary from one degree of freedom to many degrees of freedom is not well defined, and may shift with type of information desired, depending on the nature of the structure.

The analysis of impact load responses of man as part of a mechanical system ejected by a catapult was undertaken by Kroeger (1946) in the case of the interaction of a 200-pound man and a 100-pound ejection seat attaining 57 feet per second in a catapult stroke of 60 inches, for a resonant frequency of 11.1 cycles per second and a spring constant of 11,300 lbs/ft. The spring constant was derived from elastic and damping forces of seat structure, cushion, parachute pack, safety harness, etc., and of connective and muscular tissue of the body. All the variables of the catapult force were also considered. Assuming the catapult-load application to occur in one-third the period of natural frequency of 11.1 cycles per second resulted in a free response almost the same as for an instantaneous loading, and a considerable overshooting. With a rate of load application of equal to the system's period of vibration, there was very little increase of acceleration over the peak of applied catapult force.

Following up on Kroeger's calculations, Shaw and Savely (1947) plotted a family of curves from equations of motion for a linearly increasing acceleration followed by a sustained constant acceleration, representing the two portions of an idealized ejection catapult stroke. Assuming a stroke length of five feet and a final velocity of 60 feet per second as constants, the duration of linear increase in acceleration can be varied from instantaneous, with a constant acceleration of 11.2 g, to triangular up to a peak of 14.9 g at the end of the five-foot stroke. Between are combinations of increasing followed by sustained acceleration summing up to the 60 ft/sec final velocity of ejection. The rate of change of acceleration can be reduced to 100 g per second for .12 second duration with but a slight increase in the sustained portion of the acceleration from 11.2 g to almost 12 g; frequencies down to 10 cycles per second are protected at this rate of change of deceleration. A cartridge was developed by Kroeger to propel the catapult with this rate of change of acceleration. Peak accelerations of 14 g on the seat and 15 g on the subject were measured in .12 second, and a terminal velocity of 60 ft/sec was obtained without significant overshooting of g's on the subject. On the basis of theoretical and experimental work, it was found that overshooting of peak acceleration on the subject occurred at rates of change of acceleration greater than 200 g per second. Wieshofer reported three human subjects with compression fractures of vertebrae following experimental ejection accelerations of 15 g at high rates of onset. Latham and Stewart used a 110-foot ejection test tower to evaluate systematically all the factors of the ejection-acceleration problem. At the upper limits, jolt was more perceptible than peak g; Latham claimed to distinguish subjectively 40 g/sec differences in rate-of-onset. He found the rate of compression of the human trunk during ejection to depend on interaction of catapult jolt and elastic response of the seat cushion, with resultant low-frequency resonant response of the body corresponding to a system of two masses joined by springs.

This led to an investigation of whole-body natural frequencies, using a trolley with eccentric wheels up to .5-inch amplitude of oscillation, pulled along a flat floor at speeds equivalent to a range from two to 20 cycles per second. Forced resonances were found at five, eight, and 17-20 cycles per second. At eight cycles per second hip and head resonance was most evident; at other frequencies, whole body. A single impact produced by dropping the subject six inches in an ejection seat induced five cycles per second resonant response measured by an accelerometer on the hip. An impact produced by striking upward with a sledge hammer under the seat, excited oscillations of eight cycles per second on the hip and 17-20 cycles measured at the subject's head simultaneously, the latter due to head and neck response similar to a spring and mass system. Latham also used an analogue computer to analyze responses of a model system of two masses coupled by a spring subjected to the characteristics of various types of cushions, obtaining an accuracy of plus or minus one per cent.

Latham found that seat-ejection tolerance is governed by force-time function of the catapult, alignment of body and seat, and dynamic characteristics of seat cushion. For vertical acceleration of less than .2-second duration, the rate of decline of acceleration time curve after peak g is the factor that determines magnitude of overshooting acceleration on the man. Optimum duration of acceleration applied to man by an ideal cushion is .23 second to minimize overshooting measured on him. The most important variable is the low-frequency response of man, excited by the rate-of-change of acceleration. A rate exceeding five g in the first .02 to .05 second becomes subjectively uncomfortable. Tolerance is probably related to the total energy rather than to a single parameter of peak thrust.

Hess and Lombard (1958) found, from calculations of theoretical final velocities attained by a rigid body variously accelerated over a fixed distance to a prescribed maximum acceleration, that rate of rise has little effect on final velocity, which is nearly proportional to the square root of the product of peak g and stroke length for arbitrary acceleration curves. This was borne out by ejection-seat data in which peak acceleration, stroke length, and final velocities were simultaneously recorded. At most, about 10 per cent increase in final velocity can be obtained from increased rate of rise, and this could be offset if increased dynamic response to the subject should require lowering the peak g. Therefore, in the design of accelerative devices limited to a fixed stroke in which to attain a required velocity, human dynamic response can be minimized without penalizing the efficiency of the force pattern by more than 10 per cent, in terms of final velocity.

Goldman and von Gierke (1960) have reviewed all available data on human response to shock and vibration in a carefully organized almanac published in the Naval Medical Research Institute Lecture and Review Series, No. 60-3. From this report, the peak-resonance measurements on the human body have been listed in Table 1, along with the one-fourth-period durations for each frequency. For each duration and frequency, the jolt in g per second to attain peaks of 20, 30, 40, 50, 100, and 200 g, respectively, are calculated for reference to experimental measurements of single impacts in which resonant-response effects are evident.

Goldman and von Gierke note that impedance measurements for vibrations applied transversely to the long axis of the human body are not available. Data on resonant responses of suspended organ masses within the body, such as the heart, are also lacking. The effect on raising both the frequency of resonant response and the tolerance to amplitude-of-oscillation by restraints that limit displacement, such as plastic armor against the chest and abdomen, has been demonstrated but not exhaustively explored.

Bierman (1946) attempted to explore the protective value of energy absorption by undrawn nylon straps in crash restraints by dropping weighted harness assemblies on human volunteers lying supine on a table. The writer recalls seeing high-speed motion pictures of standing waves in the abdominal wall excited by the impact of a dropped harness sandwiching the subject against the table. It might be instructive to compare frequency response of the unrestrained abdominal wall with that of an abdominal wall simultaneously impacted and constricted by the abrupt, sustained load of an apron-type crash harness applied abruptly to the supine subject. At the low-frequency range for human whole-body resonance, postural tonus and orientation with respect to earth gravity for the supine, sitting, and standing positions also modify the frequency ranges of resonant response. In the standing or sitting positions, the body mass supported by the lower lumbar spine oscillates vertically with respect to the mass of the pelvis and

TABLE 1

	Peak Resonance	Period T Seconds	Jolt G per second at T/4 for peak G of					
	cps	T/4	20	30	40	50	100	200
Transverse, supine	1	.25	80	120	160	200	400	800
Whole body, unit mass	2	.125	160	240	320	400	800	1,600
Longitudinal, supine	3-3.5	.082	240	360	480	600	1,200	2,400
Vertical, sitting	4	.0625	320	480	640	800	1,600	3,200
Vertical, standing	5	.05	400	600	800	1,000	2,000	4,000
Longitudinal, abdomen	6	.041	480	720	960	1,200	2,400	4,800
Longitudinal, anterior chest	7	.035	560	840	1,120	1,400	2,800	5,600
Longitudinal, abdomen	8	.031	640	960	1,280	1,600	3,200	6,400
Longitudinal, anterior chest	9	.027	720	1,080	1,440	1,800	3,600	7,200
Longitudinal, anterior chest	10	.025	800	1,200	1,600	2,000	4,000	8,000
Longitudinal, anterior chest	11	.0225	880	1,320	1,760	2,200	4,400	8,800
Vertical, standing	12	.0207	960	1,440	1,920	2,400	4,800	9,600
Vertical, seated, head	18	.014	1,440	2,160	2,880	3,600	7,200	14,400
Vertical, standing, head	20	.0125	1,600	2,400	3,200	4,000	8,000	16,000
Vertical, standing, head	30	.081	2,400	3,600	4,800	6,000	12,000	24,000
Vertical, seated, eyeballs	60	.0041	4,800	7,200	9,600	12,000	24,000	48,000
Vertical, seated, eyeballs	90	.0027	7,200	10,800	14,400	18,000	36,000	72,000
Jaw versus head	100	.0025	8,000	12,000	16,000	20,000	40,000	80,000
Skull	300	.0008	24,000	36,000	48,000	60,000	120,000	240,000
Skull	400	.0006	32,000	48,000	64,000	80,000	160,000	320,000
Skull	600	.0004	48,000	72,000	96,000	120,000	240,000	480,000
Skull	900	.00027	72,000	108,000	144,000	180,000	360,000	720,000

lower extremities, while at the same time the upper torso oscillates transversely with respect to a hinge area in the region of the eleventh thoracic to the second lumbar vertebrae, and both these axes of response can be set in motion by either longitudinal or transverse vibrations, or by intermediate angles of application, with relative amplitudes of response depending on magnitude of the exciting component. Likewise, the head and neck can oscillate fore and aft in response to a longitudinal as well as a transverse vibration. These considerations considerably complicate the definition and analysis of resonant frequency response of the human body to impact loading, except for well-defined points of stress concentration, such as the dynamic response localized to the lower spine during catapult acceleration upward in the longitudinal axis. Compression fractures of vertebrae at 15 G, with the spine responding maximally to a high dynamic load factor, can be compared to their absence at 25 G, where the rise time is prolonged to minimize dynamic loading. Compare this with 50-G peaks tolerated in the transverse axis for the same range of application rate and duration, chiefly because the loading is not concentrated on a dynamically vulnerable elastic column. Lewis (1958) subjected himself to 26-G peak lasting 2 milliseconds at 850 G per second, measured at the lap belt, during a forward-facing seated deceleration while restrained by a lap belt only, so that the trunk was free to pivot forward. Reference to Table 1 shows that 850 G per second corresponds to a 1/4-period rise rate for the range of 6 to 11 cycles per second at peaks of 20 to 30 G's. The jolt measured at the shoulder was 328 G per second, corresponding to 1/4-period rise rate for three to four cycles per second for a peak between 20 and 30 G's. Soreness due to stretching of soft tissues was the only persistent complaint, although mild transient cardiovascular shock followed the impact. Where the trunk is held upright, an impact throwing the trunk forward against the restraining straps can be tolerated with no indication of cardiovascular shock up to 40-G peak at 331 G per second for .32-second duration, corresponding to a 1/4-period pulse of two cycles per second for 40 G, sustained for more than 1/2 the period, causing no cardiovascular shock or other adverse reaction. This was essentially a pure forced response with the body reacting as a unit mass, and body accelerations did not overshoot those measured on the seat.

Further increases of duration to peak G will not affect response, but may exceed the latent period for another order of effects. Even at rates of deceleration change within whole-body resonance frequencies, impacts attaining peaks of 40 G or more have not produced petechiae or ecchymoses about the eyes following durations of less than .2 second of forward-facing deceleration. A 45.4-G peak, attained at 494 G per second during .228-second total, resulted in scleral petechiae and retinal venular petechiae of the type described as angiopathia retinae traumatica by Purtscher in 1912, from observations on a patient whose chest was run over by a cart-wheel while he held his glottis closed; pneumatic pressure abruptly rising in the chest compressed the anterior vena cava, transmitting a wave of hydraulic pressure to retinal venules, producing hemorrhages, exudates, and edema. A combination of cardiovascular shock and Purtscher's syndrome was observed in a subject after being exposed to a total of 1.4 seconds of deceleration in the forward-facing position, which included an initial peak of 35 G applied at 600 G per second, followed by a .4-second plateau of 25-27 G, a spike of 40 G, and a final plateau decaying from 25 G to baseline in .6 second. This resulted in extreme facial congestion, confluent lateral subconjunctival hemorrhages with blebs, periocular edema, and hemorrhage.

A carotid sinus reflex induced by abrupt sustained pressure rise in the carotid arteries was subjectively observed by the writer following exposure to forward-facing deceleration of 600 G per second to a plateau of 15 G sustained for .60 second, similar to the bradycardia observed by Bierman, Wilder, and Hellems in 1946, following abrupt

chest compression by dropping a weighted undrawn nylon vest harness on a supine subject.

Apparently, these pneumatic-hydraulic effects have a latent period of less than .2 second of abruptly applied pressure to the chest.

The cardiovascular shock effect, characterized by palor, sweating, drop in blood pressure and rise in pulse rate, is related to rates of change of acceleration that correspond to dynamic loading rates for whole-body resonance, as shown by overshooting body-acceleration peaks compared to those measured on the seat or vehicle. Where the impact lasts longer than half a period of the resonant frequency, transient cardiovascular shock signs have been observed in subjects sustaining 30-G peaks or higher of accelerative force applied either front-to-back or back-to-front. Where force is applied for durations corresponding to less than half the period of resonant frequency, resulting in higher dynamic response, a lower applied peak G will incite higher overshooting peak G measured on the body, followed by rapid decline or rebound. Mild transient cardiovascular reaction has been observed at less than 25-G peaks where duration was less than .1 second; and quite severe cardiovascular shock resulted from application of 38.6 G at 1,370 G per second rate-of-onset for .12-second total duration, which corresponds to 8-9 cycles per second resonant frequency.

The severest and most persistent signs of shock were observed in a backward-facing deceleration during which Captain Eli Beeding was exposed to 2,139 G per second to a peak of 40.4 G for a duration of .040 second. This corresponds to 12-14 cycles per second as shown in Table 1. The dynamic response to this impact was a peak measured on the sternum of 82.6 G, at 3,826 G per second—approximately double the rate-of-onset and magnitude of the applied impact, which corresponds to about 12 cycles per second of resonant response. The impact force was apparently amplified with a dynamic load factor of 2.0 by the elastic response of the rib cage. Blood pressure was not discernible 30 seconds after impact, returning to 70 systolic and 40 diastolic within five minutes. Subject complained of severe lower back pain, then lost consciousness within 10 seconds after impact. Placing the subject in the supine position and elevating the legs soon restored consciousness. Subject was hospitalized three days to recover from headache and back pain, returning to duty after five days.

In the same series of experiments, a 150-pound black bear exposed to 1,779 G per second, 39-G peak for .044-second duration, had a dynamic response of 3,021 G per second and 73.1 G measured at the sternum; autopsy revealed multiple areas of congestion and hemorrhage about the ligaments and attachments of stomach, liver, and diaphragm, resulting from stretching by displacements of these organs.

DeHaven's 1942 report on survived human falls from heights of 55 to 160 feet, from which he concluded that brief impacts of 200 G were not necessarily fatal, was substantiated by experimental exposures of chimpanzees in the backward- and forward-facing seated positions to impacts of 150 G at 10,000 G per second lasting .350 seconds, by deceleration of a rocket sled from a velocity of 1,150 feet per second to a stop in 160 feet of water brakes. Stepwise increments of depth at two-foot intervals in the first 16 feet of water brakes determined the rate of increase of decelerative force.

The anesthetized chimpanzee in the forward-facing position impinged against the harness at 11,250 G per second, to a peak of 232.75 G for a total duration of .35 second. This corresponds to about 50 cycles per second rise rate. The subject survived for four hours in extreme shock. Multiple hemorrhages of heart and lungs were found

at autopsy. The chimpanzee in the backward-facing seated position sustained 16,800 G per second, to a peak of 246.85 G for a duration of .35 second, and survived with moderate injuries, shown following sacrifice for autopsy. The principal difference between the two exposures was that the chimpanzee in the forward-facing position impinged on restraint straps, elongating them several inches, and rebounded against the seat back, while the subject in the backward-facing position sustained the first impact against the backbody surface impinging on the seat and then rebounding into the harness. The indications are that impacts exceeding 200 G at high rates of application but prolonged to .35-second duration can be survived by chimpanzees optimally restrained, and that human-impact survival might fall within the same order under the same conditions.

Evans, Lissner, and Lebow (1958) reporting on drop tests of cadaver heads, found that single impacts of less than .004-second duration, attaining peaks of around 370-380 G at about 300,000 G per second, corresponding to about 400 cycles per second frequency response, did not result in fractures in two tests, but that peaks of 555 G and of 724 G, and rates of onset of 555,000 and 724,000 G per second, respectively, corresponding to about 1,000-2,000 cycles per second, did result in fractures. The first peak decayed abruptly within .002 second with small second peak where fracture occurred; impacts that did not fracture recorded a first peak that descended less steeply, followed by a fairly high second peak. Dynamic response is evident in the shape of these acceleration traces.

Any attempt at stress analysis of man with respect to impact forces must take into account the responses of the body as a whole, the simultaneous responses of different kinds, and states of materials in body structure, such as the pneumatic and hydraulic behavior of gases and fluids, plastic deformation of soft tissues, and the stretching of mesenteries and ligaments by displacement of organ masses. Since the object of human-stress analysis is to determine reversible and irreversible, disabling and fatal failure criteria for the human structure, it is well to relate measurements to points of structural weakness or of load concentration. The lower spinal column in the case of longitudinal impact force is a classic example. Resonant-response analysis applies to interaction of flexibly linked body masses, and probably to natural frequency characteristics of internally suspended organs; but factors relating to more sustained force application may determine the stress limit. Voluntary tolerance to less than 6-G amplitude of sustained low-frequency vibrations is an interesting contrast with 20-G, and even 50-G, single-impact tolerance under well-defined conditions. This alone should tend to discourage arbitrary, sweeping conclusions about tolerance, stress limits and generalized application of criteria beyond their definitions. Inescapably, the analysis will be in terms of both subjective and objective criteria having sufficient distribution of definiteness to make good medical and engineering judgment indispensable in their interpretation. This avoids the procrustean bed, on which short occupants are stretched to fit and tall ones are trimmed or compressed to the arbitrary measure. The most reliable instrument for measuring the varied effects of dynamic force on man is man.

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BIOMECHANICAL STUDIES ON THE BONES OF THE FACE

Donald F. Huelke
Department of Anatomy
University of Michigan

Introduction

Although the biomechanics of cranial-vault fractures have been thoroughly studied by Gurdjian, Lissner, Evans, and their co-workers, and these findings, as well as those studies on other parts of the body have been admirably summarized by Evans[1] in his book, "Stress and Strain in Bones," only recently have studies on the biomechanics involved in fractures of the facial skeleton been carried out.

Several years ago we began work on this problem wherein the use of basic engineering tools of non-destructive testing procedures (Stresscoat and strain gauges) and destructive testings involving fracture-producing impacts on individual bones and intact heads have been employed. Thus far, concentration of effort has been placed on the lower jaw, with work on the mid-face region in the early stages.

Results

Preliminary findings indicate that the bones of the face fracture due to tensile failure[2]. The bones are being stretched and fail due to tension, not to compression. In the lower jaw the following, in general, can be expected: failure (fracture) of the jaw at the point of impact and at the narrow area of the mandible—the subcondylar area—the region just below the temporomandibular joint. These generalizations agree with and reinforce the findings on other body parts—the brain case, thigh bones, vertebral column and pelvis—that fractures are produced due to high tensile strain within the bone.

One cannot reinforce these body structures from the inside, but protection must come from without. To do this the amount of energy absorbed by the facial bones must be decreased. This can be done by protecting the face by some sort of energy-dissipating "mask," by restraining the head and body to prevent impact against objects, or by covering possible impact surfaces and objects with energy-dissipating material. Each of these methods of protection has its own application and rarely will all three be simultaneously used.

Problems Unsolved

Along the present lines of investigation certain outstanding questions need to be answered: (1) What is the modifying effect of the teeth when in occlusion (biting together) at the time of impact? (2) How do impacts to the eyeball produce the blow-out fractures of the eye socket? (3) What is the microscopic anatomy of the thin, yet strong, bone of the mid-face area?

An entire area yet untouched is the biomechanical characteristics of skin: (4) What are the energy-dissipating qualities of human skin and underlying soft tissue? (5) How does skin tear? (6) Does this differ with the type of impactor—a blunt instrument versus a knife edge? These and many other questions with regard to skin and soft tissue need to be answered.

In conclusion, further investigations into the facial region as to the mechanics of both bone and soft tissue, their fracturing characteristics, and level of energy absorption need to be studied. Such data are necessary for the design engineer to prepare proper protection for the individual.

Facilities Available

At the Wayne State University, Colleges of Engineering Medicine, complete biomechanics testing laboratories are in operation. Both non-destructive and destructive testing procedures can be carried out, and most all necessary equipment for recording such tests is available. At The University of Michigan, Department of Anatomy, a complete and functional Stresscoat laboratory for non-destructive testing is operational. Additionally, adequate statistical analyses of data can be carried out here. Furth material for study is available in the department of anatomy at both medical schools.

Research Personnel

Lissner, H. R., M.S., Professor, Wayne State University, Department of Engineering Mechanics.

Patrick, L. M., M.S., Associate Professor, Wayne State University, Department of Engineering Mechanics.

Huelke, D. F., Ph.D., Assistant Professor, The University of Michigan, Department of Anatomy.

Lange, Wm. F., M.D., Plastic Surgeon, Detroit, Michigan.

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"Deformation Studies of the Bones of the Face," Grant D-1416 USPHS, National Institute of Dental Research. Principal Investigators—H. R. Lissner and L. M. Patrick, Department of Engineering Mechanics, Wayne State University.

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STEERING WHEEL IMPACT*

Murray Burnstine
Assistant in Legal Medicine
Harvard University

The object of this paper is to describe briefly the manner in which various automobile steering wheels have been observed to function as either injury-producing or protective structures. The steering wheels were first studied at the scene of fatal collisions, where their deformation and deviation from installed position was recorded. The steering wheels and often the complete steering wheel-mast jacket assemblies were subsequently removed for more detailed laboratory analysis. To date, approximately 40 steering wheels have been accumulated and correlation of steering-wheel deformation with injury data from hospital and post-mortem findings is in process. Lacking a clinical "proving ground" to evaluate design proposals, these observations might be of value to the automotive-design engineer.

Many totally demolished automobiles have been examined which yielded no evidence of occupant-steering-wheel impact. It is only under a unique set of conditions that the steering wheel can function as a decelerating device. Figure 1 illustrates the

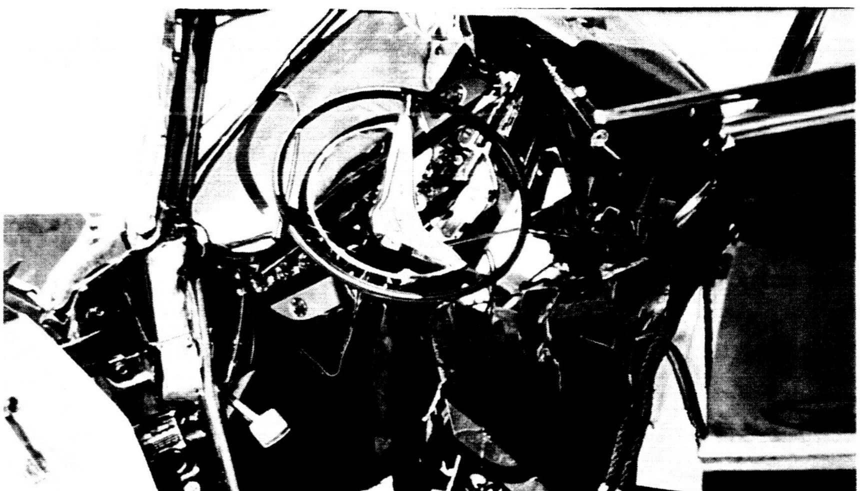

Figure 1

*From the Department of Legal Medicine, Harvard Medical School, Research carried on under Grant RG-6084, U.S. Public Health Service, Department of Health, Education and Welfare.

extreme case of a right-angle, right-side vehicle impact. All three front-seat occupants received fatal injuries. Several cloth fibers were found imbedded in the measurably undeformed steering wheel. Injuries related to interior vehicle structures other than the steering wheel indicated the occupants travelled directly to the right, the driver brushing across the steering wheel. The steering wheel is similarly not deformed in vehicle non-collision roll-over situations.

Although Mr. Moseley's paper(1) includes a more detailed analysis of occupant movements, the following graphical method of reconstruction is useful in determining the attitude at which occupants impact the steering wheel. In general, lines drawn on a plan-view sketch of the vehicle, originating at the point of maximum vehicle deformation and radiating to the seated positions of occupants, will closely describe occupant path of travel during vehicle deceleration. The above method of reconstructing collision events has proved compatible with deformation and observed related injuries. When seat backs do not collapse the method is valid for occupant rebound following rear-end collisions. In a similar manner, a vehicle impacted at the left-front or right-front will result in a diagonal path of occupant travel so that passengers rather than the driver can impact the steering wheel rim.

With respect to the driver, the steering wheel can function most effectively as an arresting device only under the condition of near-central-frontal vehicle collision. Unfortunately, frontal deformation, combined with the present vehicle design practice utilizing a long rigid steering-gear shaft within a mast jacket clamped to the instrument panel, allows the steering wheel assembly to alter its orientation with respect to the driver. Reconstruction of the time sequence of collision events through correlation of observed injuries and related vehicle structures indicates that considerable deformation occurs while occupants are at rest. Steering-wheel structures in their deviated positions are then impacted by occupants. Several modes of steering-wheel motion have been observed. Figure 2 illustrates the most frequently observed sequence. Vehicle sustained head-on impact with the vehicle shown in Figure 3. Maximum deformation of forward structures was located to the left of vehicle center. Combined radial and axial loads at steering-gear end of steering shaft forced mast jacket-wheel assembly into passenger compartment, shearing under-dash bracket bolts. Subsequent injury picture indicated that both the driver and the middle front-seat occupant, who expired, impacted the wheel assembly after it had been forced to the left and downward as shown. Both front-seat occupants of the other vehicle, shown in Figure 3, survived.

The recessed hub-design wheel (Figure 3) proved effective in decelerating both the driver who impacted the lower left wheel-rim quadrant and the passenger who, because of the recessed-hub design, was able to impact the lower right spoke rather than radially into the higher rate rim. The steering wheel could not have functioned as it did if it had deviated from its design location. A recessed-hub wheel impacted in design location might have reduced injuries to a survivable level in the vehicle shown in Figure 2.

The vehicle shown in Figure 4 struck a concrete bridge pier head-on. Driver and front-seat passenger were killed by forward vehicle structures entering the passenger compartment. The illustration serves to demonstrate the possible extremes in angular deviation from installed positions of mast jackets. The under-dash bracket has not failed but acted as a hinge, allowing the mast jacket to assume an almost vertical position. In Figure 4 the steering wheel has been removed from its final position in contact with the head-lining material.

Figure 2

Figure 3

Figure 4

The steering wheel shown in Figure 5 was removed from a vehicle which struck a cinder-block building. Low frontal vehicle structures were involved with resultant pitching of the driver radially into lower half of steering-wheel rim, illustrating the necessity of radial-energy absorption potential as a design parameter.

The recessed-hub steering-wheel design functions as a load-limiting device while yielding. Its effectiveness is due in part to the ability of the human body to deform and distribute the total load among adjacent structures. When the load application is through the relatively large-circumference rim as compared to the hub, the resultant shearing stress is reduced considerably. Bruising due to local high unit bearing or contact pressures can be reduced by increasing the projected area of the steering-wheel rim. Most of the energy-absorption characteristics of the wheel assembly are dependent on spoke design, and a thicker rim would reduce contact pressures for both the axial and radial impact modes.

Laboratory evaluation of prototype design possibilities must of necessity involve suddenly applied or impact-type test fixtures. Cadavers instrumented to generate

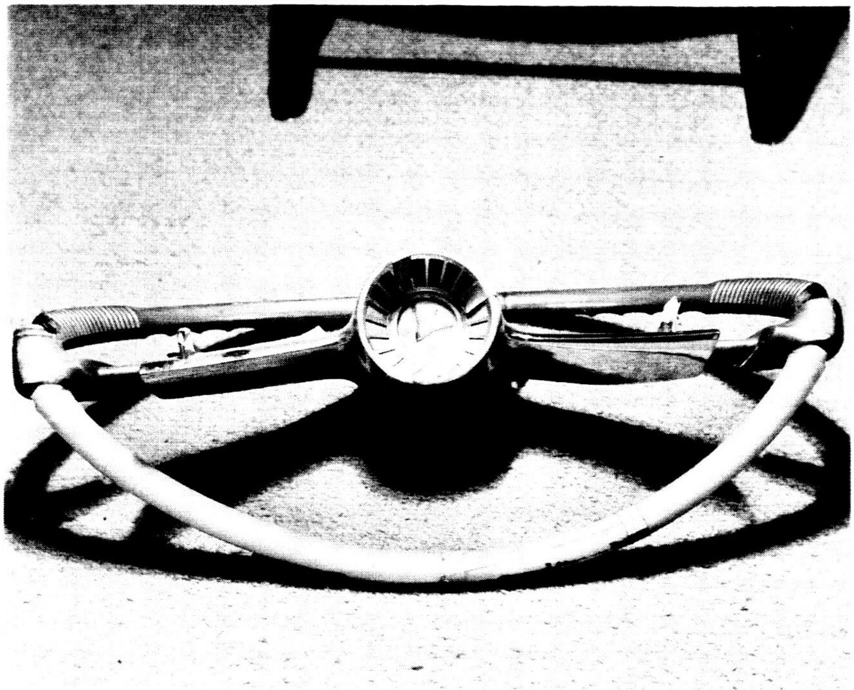

Figure 5

deceleration data correlated with bone-structure damage from x-ray films and autopsy findings would yield the most useful data. The importance of impact testing of steering wheels can be substantiated by examples of impact failure observed in the field. A six-foot length of 3 x 5-inch timber guardrail was split longitudinally after piercing the firewall and impacting the steering-wheel rim. Several rim-to-spoke separations have been observed, without sufficient deformation to be associated with gradually applied load phenomenon.

The horn ring cannot be ignored as an injury-producing structure. It is not capable of energy absorption and shatters upon impact. The resultant exposed sharp edges serve only to identify the driver. Drivers wishing to remain anonymous usually purchase the minimum-trim body style, which features the less lethal horn button of 30 years ago.

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12860

CRITERIA FOR INJURY POTENTIAL

Charles W. Gadd
Research Laboratories
General Motors Corporation

Example of Laboratory Evaluation

Appraisal of injury potential of automotive instrument-panel paddings, visors, projections, and the like may be attempted at any of a number of levels of sophistication. The first might be to select (considering a crushable material) on the basis of its load-deflection properties as indicated in A and B of Figure 1, obtained at some arbitrary rate of loading and unloading. Choice on this criterion would be in favor of B, the material having greater energy-absorption capacity. Resistance to deflection of padding materials is, however, generally a function of rate-of-loading, with the result that the hysteresis loop narrows and shifts in the direction of curve C. Rate of loading is thus immediately seen to be a factor bearing upon the suitability of the material. This factor may be taken into account only by selecting test impacting hammer velocity or velocities most representative of actual accidents, for which one must resort to statistics of accidents as well as studies of car and occupant kinematics in the accident. The latter will entail theoretical calculations as well as high-speed-film analysis of staged accidents.

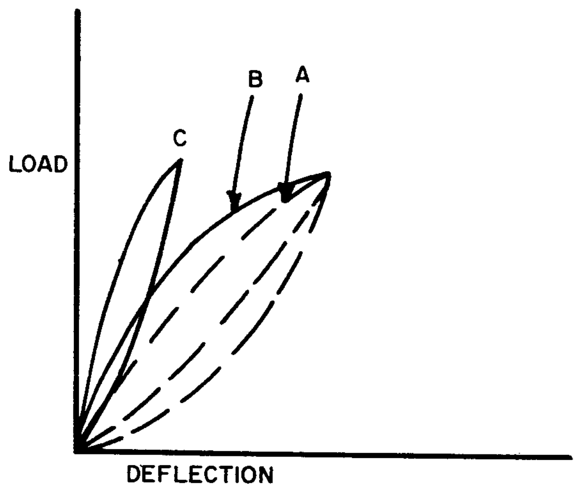

Figure 1. Load—deflection curves for padding materials.

A third level of refinement in test procedure takes into account still another important factor—the yielding and energy absorption of the underlying or surrounding structure. Instrument-panel and visor studies, for example, have shown that energy-absorption capacity of the underlying sheet-metal structure may be 10 times that of the absorption of the padding itself. For this reason, it has been necessary in test fixtures such as that of Figure 3 to incorporate a full-scale section of the surrounding body capable of yielding as it would in the accident.

Interpretation of Test Results

This presents problems with which the investigator can only partially cope at the present time. While deceleration-time or force-time curves are readily obtained as oscillographic recordings of the impact, these cannot be interpreted in terms of probable trauma for the great majority of cases in which the latter is relatively superficial

in nature. Instead, in the writer's laboratory, it has generally proven more fruitful in such cases to obtain an "integrated" estimate of injury potential by observation of damage to an expendable specimen of animal skin and underlying tissue applied to the impacting hammer before each test. Selected to simulate human tissue as closely as possible, this is believed to give the most valid possible measure of superficial injury. Deeper injury, as, for example, to the knee, is also estimated by inclusion in the forward portion of the test hammer, under the animal tissue, of an expendable section of end-grain wood having crush resistance similar to that of average bone.

Criteria for Deep-Seated Injury

Still more deep-seated injury, for example, brain concussion, must also be considered. This, of course, does not lend itself to appraisal by observation of degree of superficial injury, and new criteria must be developed. An early indication of the direction which such research should take was given by Lissner and Gurdjian in experiments which showed that various combinations of cranial pressure and times of exposure (to these pressures) necessary to produce severe concussion could be represented by a curve having the form shown in the upper part of Figure 2.

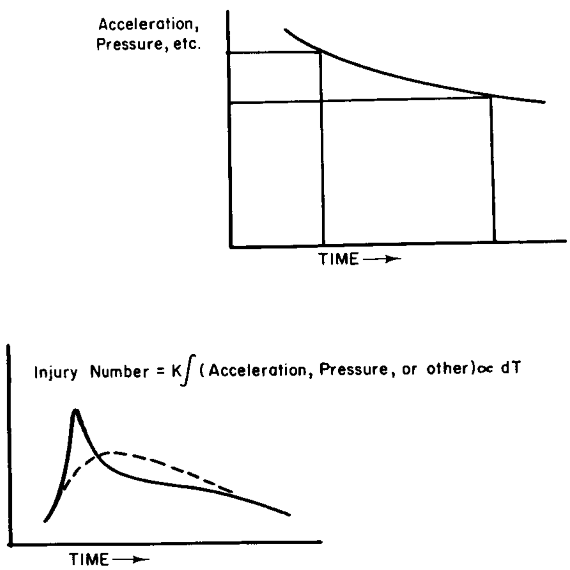

Figure 2. Empirical injury criterion compatible with tolerance curves of type shown at top, but capable of handling variations in impulse profiles as shown below.

The general form of this curve has been substantiated by other investigators, most of whom have employed acceleration rather than pressure as the ordinate, and it appears to be fairly consistent, even for differing modes of injury and over a variety of types of animals, except for the expected higher numerical value of acceleration tolerance for smaller scale animal subjects.[1,2] Best fit of data to a straight-line injury-tolerance curve will normally be obtained if the loading as well as time-duration of loading are plotted logarithmically.

This type of presentation is however, restricted to a single-impulse wave-form (square or trapezoidal for example), and for this reason cannot handle questions of relative injury potential to be expected from differing wave-forms or profiles as shown in the lower part of Figure 2, which might result, for example, from striking the head against two structures having totally different crush characteristics. In two instances such as these the average value of the ordinate might be the same for each impulse, whereas true injury potential would reasonably be expected to differ.

As a means of approximately evaluating differing-impulse wave-forms such as these, the following "injury number" is suggested:

$$\text{Injury or damage} = k \int (\text{acceleration, pressure, etc.})^{\alpha}\, dt.$$

This is obtained by integrating the imposed loading, whether it be acceleration, pressure, or some other measurable quantity, over the entire profile of the impulse. The exponent α is a weighting factor expressing the greater dependence of accumulated damage upon intensity of loading than upon time-of-exposure to that loading. If the exponent is set equal to one, borderline injury curves of the type shown in the upper part of Figure 2 reduce to a straight line at 45 degrees on log-log paper, and the criterion reduces to a statement that damage or injury should be proportional to the area under the impulse profile or to the average value of loading. This is a useful approximation at present, but by analogy with failure properties of engineering materials, an exponent greater than one should be expected, to take account of the fact that the highest-level portions of the profile should contribute by a disproportionately great degree to the total injury.

The numerical value of the above exponent α should vary with the time-dependency and mode of failure of the critically strained elements of tissue involved. In predicting failure of a uniformly strained or stressed specimen of engineering material, the part played by time-of-exposure to loading may be taken into account quantitatively through knowledge of the viscoelastic properties of the material and its strength under its possible modes of failure. When studying the more complex situations typical of tissue failure, on the other hand, rigorous analysis is not possible and an overall empirical function may be the best representation available.

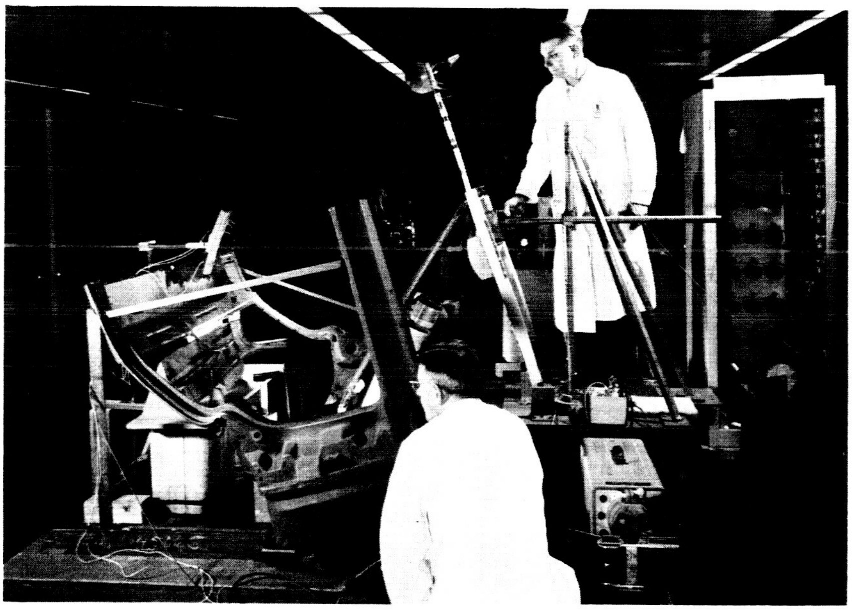

Figure 3. Fixture for Laboratory impact testing of Body Interior paddings and components.

Failure of Complex Structures

The foregoing criterion, while apparently roughly applicable throughout the body and over a wide range of time-duration of exposures to loading, should, strictly speaking, be true only for uniformly loaded material free of resonant or forced distortions. Deviation from Figure 2 thus should occur in more complex structures when more or less discreet masses and elasticities exist which can react dynamically. A resonant mode made up of these can then "tune in" with pulse width, or rise or fall time of the pulse, and aggravate tissue strain and resultant injury.

This phenomenon was studied, again taking an analogy from engineering experience, in earlier days of the automotive industry when the effects of combustion-pressure wave-form, and in particular of rate-of-pressure-rise, upon roughness of operation of the engine were a matter of much controversy. This problem was fully resolved and understood only when the pertinent structural modes of the engine were reduced to lumped-parameter systems and the reactions of these systems to various pressure wave-forms calculated by the use of numerical integration methods.[3]

In Figure 4 is shown a series of calculations made at that time which is typical of results obtained from this approach. For a given impulsive excitation (here triangular in shape), the amplitude of structural resonance is seen to be greatest if its

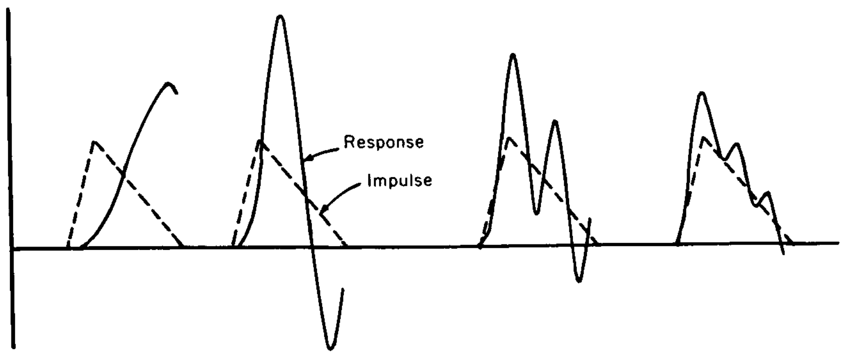

Figure 4. Displacement response of mass-spring system of varying natural frequency to a given impulse.

half-period roughly corresponds in time duration with that of the pulse. This is true of the second example in Figure 4. High rate of pressure rise or "onset" was found to play an important role only if structural resonances were correspondingly high in natural frequency.

Aside from transient resonant effects, a second and more important departure from the simple weighted-impulse criterion occurs when the body is subjected to severe distortional forces. Under these conditions, concentrations of stress occur, each of which must be studied individually.

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12-861

EXPERIENCES IN HEAD INJURY AND SKELETAL RESEARCH

E. S. Gurdjian
College of Medicine
Wayne State University

Impact studies have been conducted at Wayne State University for the past 20 years with the cooperation of the departments of neurosurgery, anatomy, and engineering mechanics. While the majority of these studies involved impacts to the head, other areas of the skeleton were also studied. In this report, a brief summary will be given of the research that has been conducted in the past by our group and an outline of the future studies for impact to various parts of the human body.

Impact Studies Involving the Head

Tests were conducted with human cadaver skulls with the aid of stresscoat brittle lacquer, to determine the mechanism of fracture and fracture locations due to blows delivered to various parts of the head. The results of the study indicated that a blow to the head produced a localizing bending at the point of impact and an undulating in-and-out bending took place about the point of impact at some distance from the location of the blow[1, 2]. Of course, depending on the velocity of the injuring energy the area of impact would fail and a perforation or depression of this region could result[3, 4, 5]. In the absence of failure of the skull at the point of impact, the area of impact was inbended, and outbending occurred at a considerable distance from a point at which the blow was struck. It was learned that the skull generally failed due to the outbending associated with tensile stress on the outer surface of the skull, and the break was initiated at a considerable distance from the point of the blow and progressed toward the point of impact and in the opposite direction. This is the mechanism of the production of a linear skull fracture (Figures 1, 2, 3).

As a result of about 1,000 tests to 100 skulls selected at random, an atlas was prepared showing the region in which the initial linear fracture was to be expected due to a blow in any area of the skull.

The energy required to produce a skull fracture was determined. Tests were performed on dry skulls as well as on intact cadaver heads, and it was found that about 10 times the energy was required to fracture the skull when the intact head was used as was necessary to fracture the dry skull, the additional energy being absorbed by the scalp when the intact head was tested. In 60 heads this indicated that the energy necessary to produce fracture to the skull varied from about 400 to 1,000 inch-pounds, with an average of about 600 inch-pounds. While some difference was found in the energy required for fracture due to blows in different locations, the tests were not extensive enough to determine that these values were significant. In other words, greater variations were found in energy requirements for fracture with a blow in any single location on different skulls than were found from one position to another on the same skull.

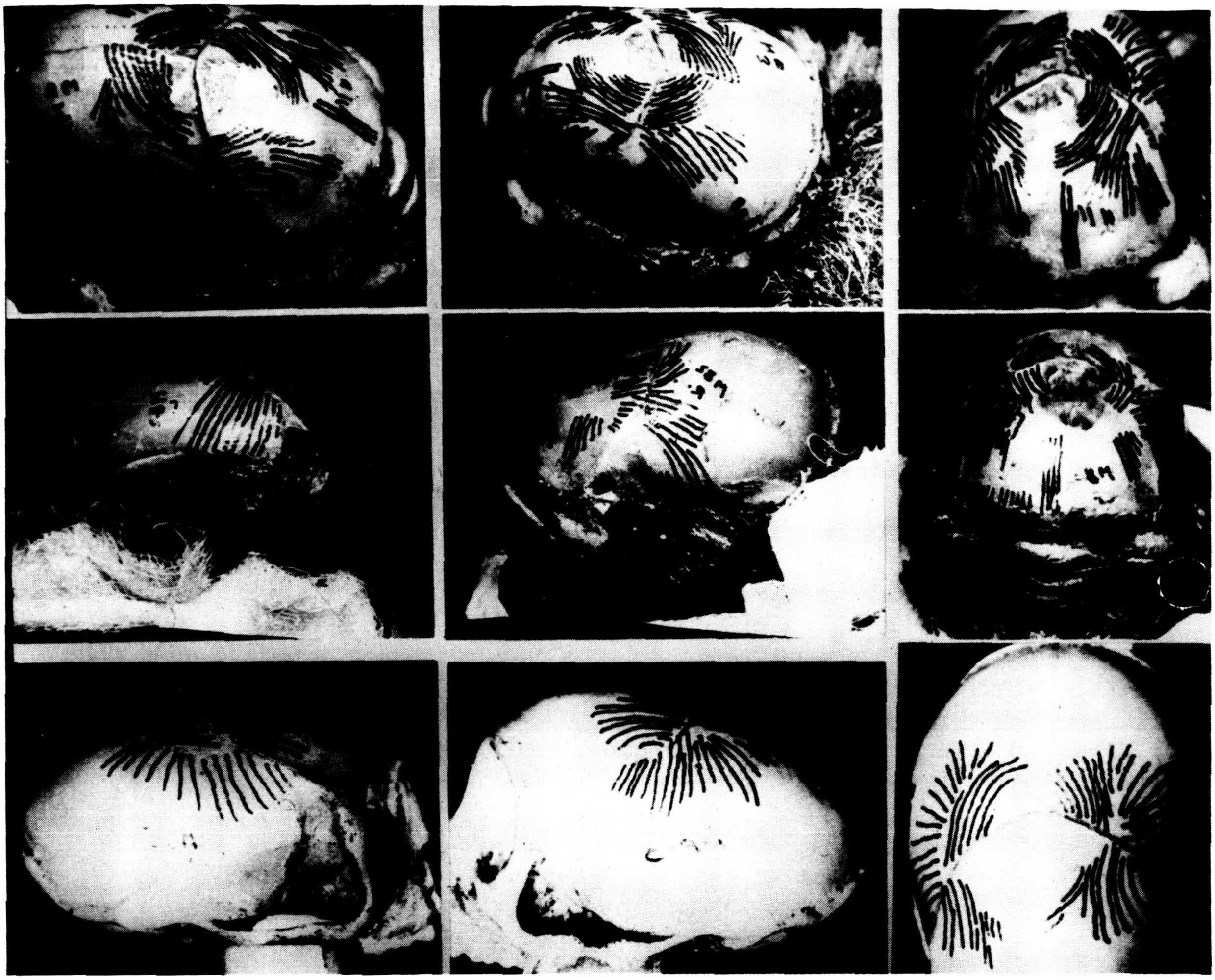


Figure 1. Stresscoat studies. Stress patterns of the skull following a mid-parietal impact with the rhesus monkey are essentially the same whether the animal is under anesthesia, as shown in the upper row, the animal has been sacrificed with contents of the skull intact, as shown in the middle row, or in the dry skull of the same animal, shown in the lower row. That one can use the dry skull of an animal to study the deformation patterns to obtain information on the patterns in the living animal, by this technique, is a valid conclusion from the above experimental data.

Time during which the energy was absorbed in order to produce a fracture was determined in the following manner: A small portion of the scalp was removed in the area in which the fracture was expected due to the position in which the blow was to be delivered. An electric strain gauge was cemented to the skull across the direction of the expected fracture. The time was recorded from the initial contact of the head with a flat surface upon which it was dropped to the opening of the electrical circuit of the strain gauge as the crack, or fracture, passed through it. In this type of testing, it was determined that the scalp absorbed energy for approximately the first 6/10 of a millisecond, followed by the build-up of strain in the skull for the next 6/10 of a millisecond, at which time the fracture occurred (Figures 4 and 5).

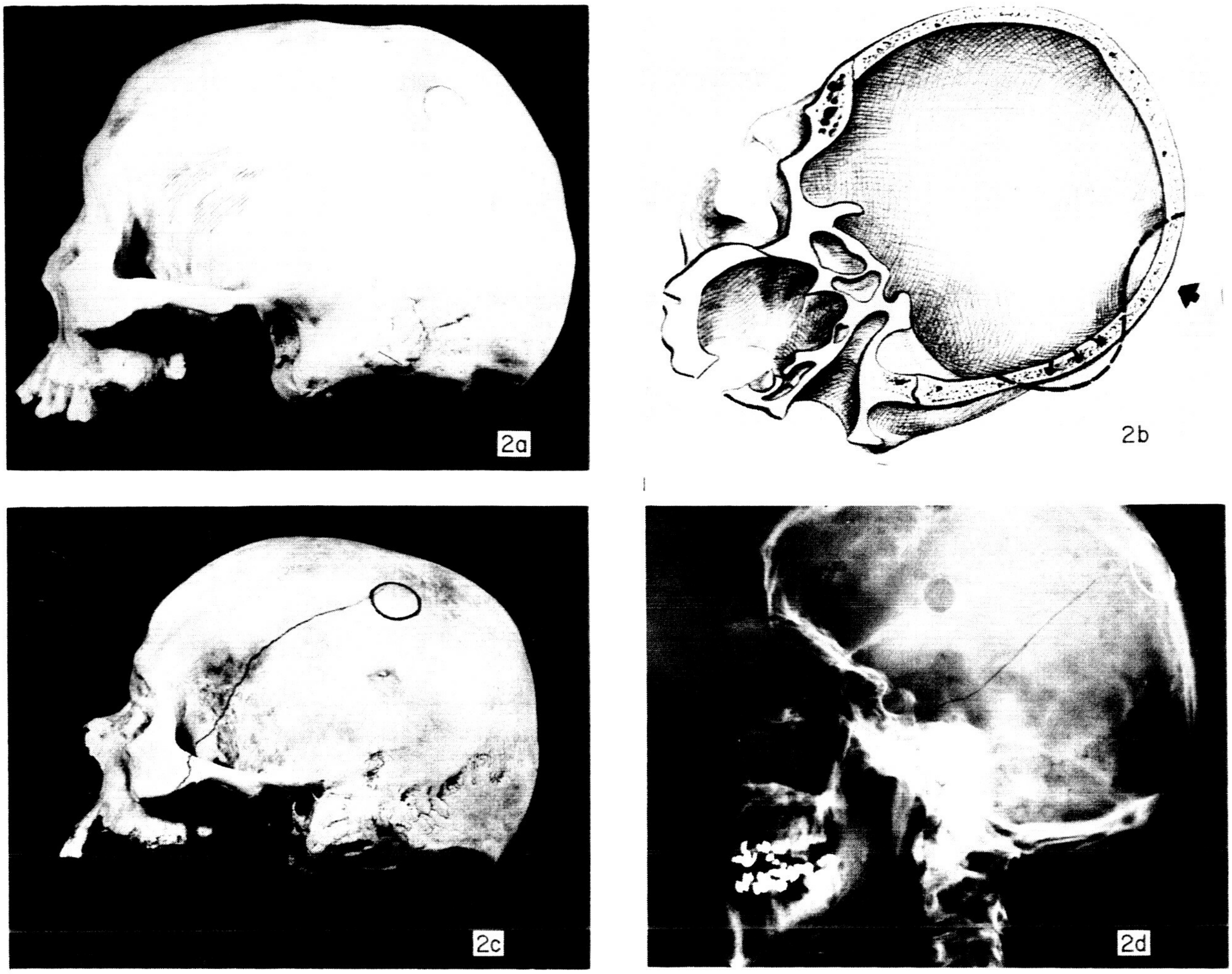


Figure 2a, b, c, d. "A" demonstrates the tensile forces in the temporal region in a human skull after impact in the parietal area, shown in the circle. "B" shows the deformation pattern when there is no failure of the bone at the point of impact. The area of impact is inbended and distal to this area; the skull is outbended, shown by the arrows. It is in the outbended portion that a linear fracture is initiated due to tensile stresses. The fracture thence extends toward the point of impact and toward the base of the skull. "C" shows an experimental skull fracture in the cadaver following impact in the parietal area shown by the circle. Note that the linear fracture is wider in the temporal region than at the point of impact. There is also a fracture of the zygomatic arch. The fracture was initiated in the temporal area; thence it extended toward the point of impact and toward the base of the skull. In "d," the point of impact in this patient is shown by the area of laceration, marked with a circular wire. Fracture resulting from the impact due to a fall is shown extending toward the temporal region. Note that the fracture line is much wider away from the point of impact than at the point of impact. These four portions of Figure 2 illustrate the mechanism of a linear skull fracture.

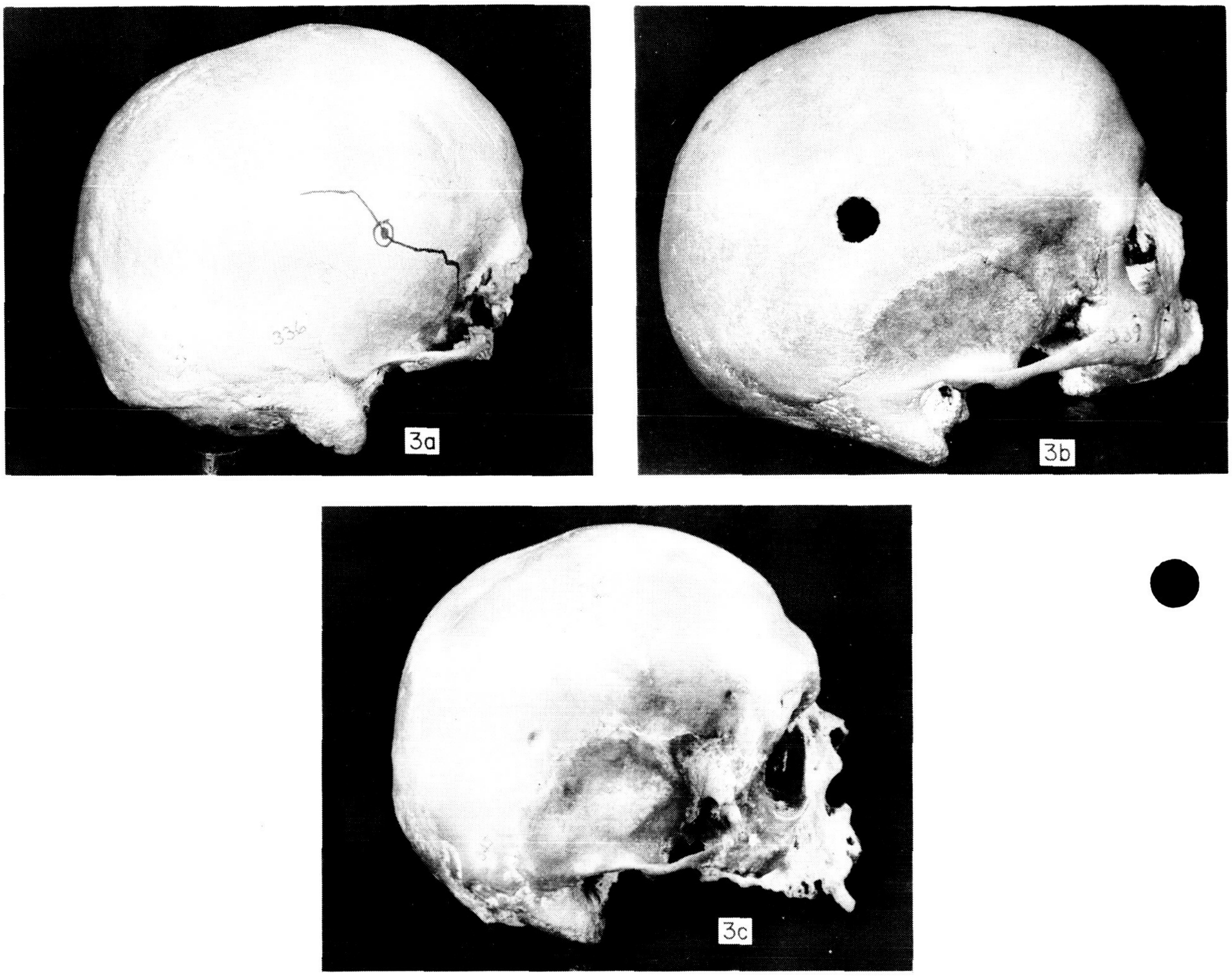


Figure 3a, b, c. The velocity of the injuring object is important in the type of fracture produced. A steel ball one inch in diameter at an impact velocity of 50 feet per second has caused a linear fracture in the temporal area as shown in "a." The area of impact is shown by the inked circle. The same steel ball dropped from a higher point with an impact velocity of 90 feet per second makes a clean hole through the skull as shown in "b." In "c," note that when a steel ball half the size of the previous one is dropped at an impact velocity of 90 feet per second on the skull, there is an inbending only of the outer table of the skull. The kinetic energy of the injuring object was not sufficiently high to cause a perforation here.

With strain gauges cemented to cadaver skulls, natural frequency of the skull was determined by impacts delivered to the head. This was found to be in the neighborhood of 700 cycles per second[3].

Many tests were performed in order to determine the strength characteristics of the bone itself. These tests were conducted on bone freshly obtained from amputation, or on dry bone and bone that had been maintained in embalming fluid for six months to a year. Adjacent samples of bone were taken in these tests to determine the effects of embalming on the strength of the bone. Samples of bone were also obtained from different parts of the same bone to determine strength variations from different regions.

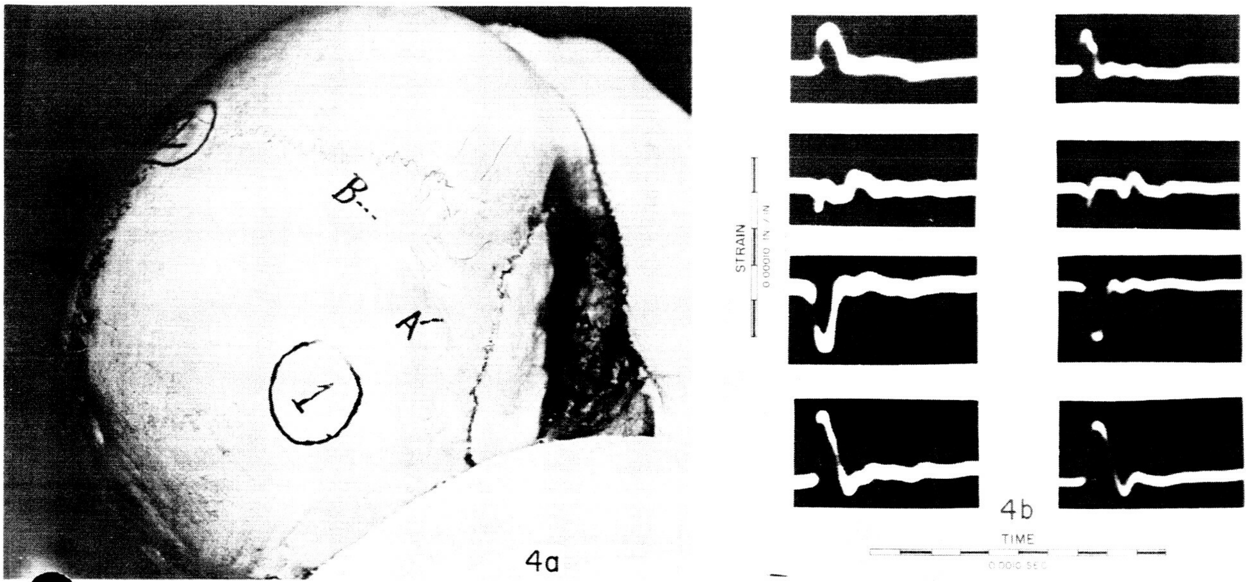


Figure 4a, b. "A" demonstrates the study of the deformation patterns of the human skull soon after death by the use of electric strain gauges. In "b," the deformation pattern of the skull studied by this technique is shown. The entire disturbance lasts about 4/1000 second at the rate of 700 cycles per second. The greatest amount of deformation occurs in 6/10,000 second after impact.

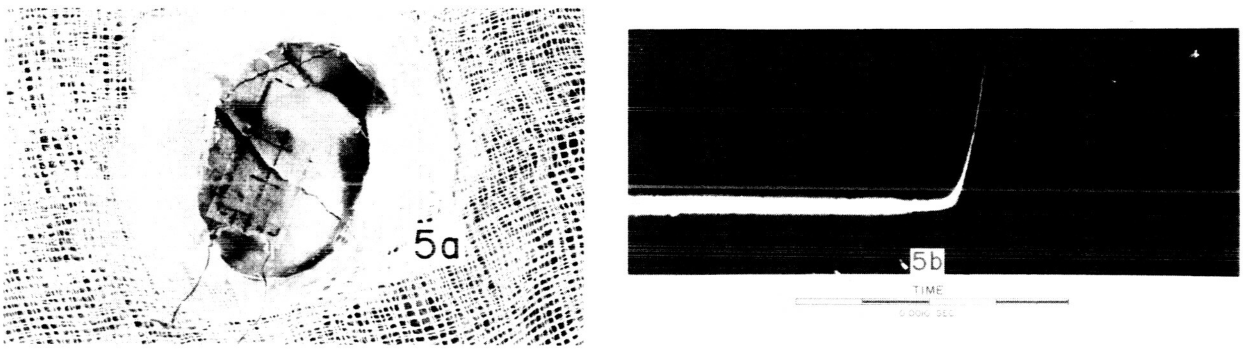
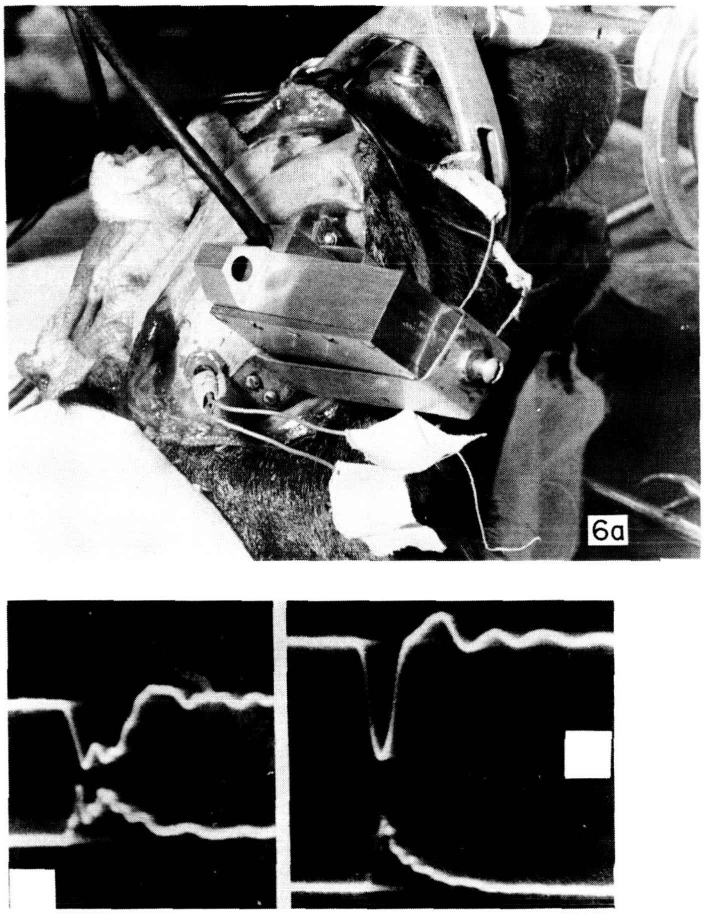


Figure 5a, b. Time of linear fracture following impact is shown in "a." A dual-beam oscilloscope records the time of impact and the exact time of fracture resulting in opening of the electric circuit. Note that a small mass of luminous matter just below the thick line indicates time of impact; the end of the upgoing line indicates the opening of the circuit or the time of fracture. Between the time of impact and beginning of the deformation of the skull there elapsed 6/10,000 second, and from time of the beginning of the deformation of the skull to the time of the opening of the circuit there was another 6/10,000 second, at which time the skull fractured. In "b," the actual tear through skull and strain gauge is shown.

From these tests it was determined that embalmed bone was generally somewhat stronger than fresh bone as tested directly after being amputated, but variations in the strength of the bone from one individual to another and from one bone in the body to another in the same individual were as great as the variations between the fresh and the embalmed bone. Tests were also conducted to determine the strength and the modulus of elasticity of spongy bone. From these studies it was concluded that tests made on bone in cadavers that had been embalmed would give as reasonable results as if the bone were fresh.

A very large portion of our research was concerned with mechanism of production of concussion due to a blow to a head. Dogs were used principally for these studies, and accelerations of the head were measured as well as the pressures within the cranial cavity (Figures 6 and 7). Pressure determinations were obtained by screwing



6b

Figure 6a, b. "A" shows experimental set-up for recording intracranial pressure by the screwed-in pressure gauge and the acceleration with the Statham accelerometer, in the dog under anesthesia. Impacts were delivered to the left parietal area, the head supported in the hand. In "b," typical records of intracranial pressure (lower record) and acceleration (upper record) are shown.

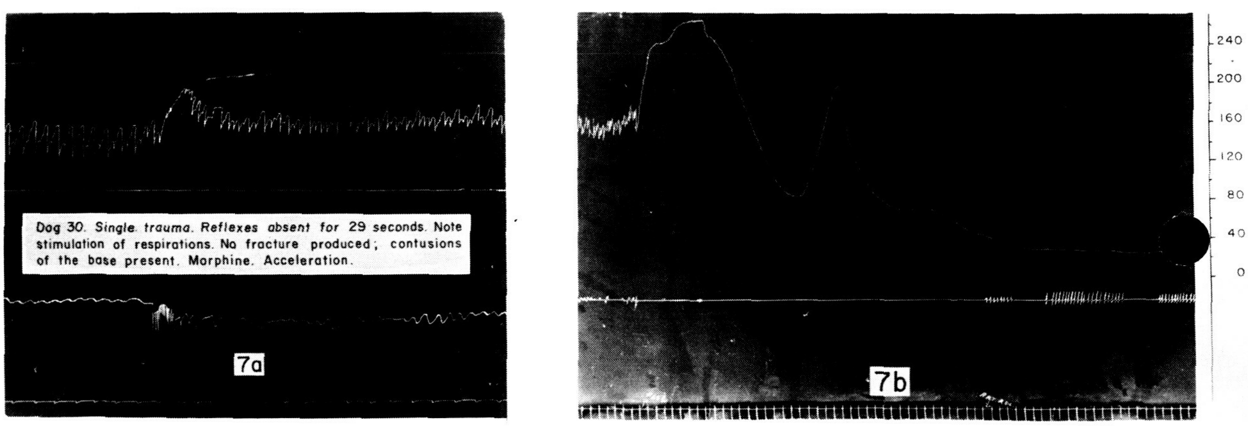


Figure 7a, b. Records of blood pressure and respirations in the experimental animal showing a minimal concussive effect in "a" and a severe concussive effect in "b." In "a," the animal survived; in "b," the animal died.

pressure pickups through the skull and having the sensitive element of the pickup in contact with the cerebrospinal fluid. Accelerations and pressures produced by impacts upon the head, and their time durations, were studied. Sudden increases in pressure at the time of impact appeared to be a function of the volume-change due to the bending of the skull, and were also due in part to the acceleration imparted to the head. Blows to the head of the dog in many instances produced local fracture at the point of impact, with resultant brain damage which obscured the concussive effect. In order to eliminate this variable from the experiment, pressures were applied directly to the brain by means of a valve which controlled the magnitude of the pressure applied and its time duration. From these tests it was learned that both the magnitude of the pressure and the time of its duration were important factors in the production of concussion. Concussion could be produced with a high pressure lasting a very short time (90 psi for one millisecond), and the same concussive effect could be produced by low pressure lasting for a longer time (20 psi for 10 milliseconds). The pressure-time relationship, however, was found to be nonlinear, and, therefore, an explanation other than impulse, which is the product of pressure and time, was sought. Due to the presence of the foramen magnum at the junction of the skull, neck, and spine, a shear stress is developed in the brain-stem area whenever a pressure is built up within the cranial cavity. The magnitude of the shear stress is a function of both the magnitude of the pressure and its time duration. In order to confirm this theory, pathological examination of the brain was undertaken and cell damage was localized in the brain-stem area where shear should be developed. Microscopic examination of the tissue showed evidences of cell damage in the brain-stem area and in the region of the base of the brain[6]. No evidence of damage to the tissue, even in the area where the pressure was applied to the dura, was found. Further confirmation of the presence of the shear stress was obtained by constructing a plastic model, filling it with milling yellow which becomes doubly refracting in a polariscope when subjected to shear stress[7]. The model was struck and the resultant pattern in the milling yellow was obtained with high-speed cinephotography. This demonstration confirmed the presence of shear stress in the brain-stem region due to a blow applied to the surface of the skull (Figures 8 and 9).

Impact Studies of Femur, Pelvis, and Spine

Longitudinal, torsional, and bending impact tests have been conducted on the femur (Figures 10 and 11). In some tests, the femur was coated with stresscoat, and the pattern of cracks in the stresscoat indicated the weakest regions where fracture would first occur. Other tests were conducted at energies sufficiently high to produce fracture, and these tests confirmed the indications obtained with the stresscoat tests in regard to the location of the fracture. Production of fracture of the femur was also studied, using the intact leg. In this case, an incision was made to expose the bone and a weight was dropped on the head of the femur. High-speed photography, 2,000 frames per second, was used in order to watch the behavior of the femur due to the blow and the initiation of the fracture.

The effects of impacts to the pelvis and spine were determined by dropping the pelvis and spine onto a large steel block. Stresscoat was previously applied, and the cracks in the stresscoat were studied to determine the deformation of the pelvis due to the blow. Since the proper distribution of weight was not present along the spine in these tests, further tests were made using the intact cadaver and exposing a portion of the pelvis which was stresscoated. The entire trunk of the cadaver was then dropped on the pelvis on the ischial tuberocities, and the pattern obtained in the stresscoat was used to determine the type of deformation occurring and to predict the location of

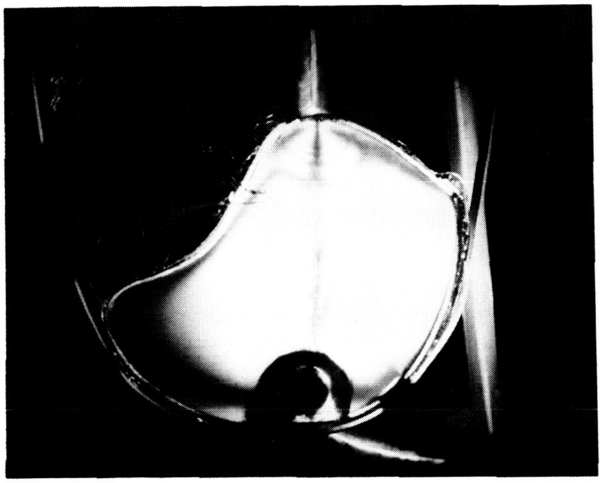


Figure 8. Photoelastic analog representing a mid-sagittal segment of the human skull and spinal canal junction, in an inverted position. Impacts upon the side, front, or back of the plastic flask resulted in shear forces at the simulated craniospinal junction. The degree of the shear is proportional to the number of bands. In closed-head injury with no perforation or failure of bone, shear forces may occur in the craniospinal junction, resulting in the unconscious state characterizing a concussive effect.

fractures. Tests are currently being performed in which controlled accelerations are applied to the entire cadaver through the ischial tuberocities. These tests are conducted with the use of our accelerator, which will provide a maximum acceleration of 50 g to an entire cadaver for a period of 1/10 second. Provision is made in the accelerator to vary the shape of the acceleration pulse, the rate-of-onset of the acceleration can be varied from 50 g to over 2,000 g per second. Of course, at the lower rates-of-onset we cannot reach the higher g values since an eight foot stroke is the maximum that we have available.

Tests conducted on the cadaver have consisted of the application of strain gauges to the spine at various locations. The strain-gauge output is calibrated through the application of dead loads through the head and shoulders of the cadaver. The strain-gauge readings can then be interpreted in terms of pounds of force occurring at the various vertebrae on which the gauges are attached. The effect of various types of restraint on the cadaver have been studied by this technique.

Current investigation is determining the effect of varying the rate at which the acceleration is being applied to the cadaver. Other studies on the spine have been made to determing the effect of applying repeated loads both in compression and bend- in the attempt to determine the mechanics of disk herniation and protrusion. Loads in some instances have been applied as many as one million times, and the only herniations produced have been into the bodies of the vertebrae and not into the spinal canal. Further testing in this area is contemplated, with the addition of torsional loads to the other types which have been used in the past.

Recently, we have studied the effect of cadaver-head impact into several types of safety glass, in order to determine the severity of impacts (Figures 12 and 13). Both laminated and tempered glass were used. This evaluation was possible because of the application of the theory of concussion which we have proposed. Clinical experience has shown that impact to a hard, flat surface frequently produces a linear fracture and a mild-to-moderate concussion simultaneously. By dropping the cadaver so the head struck a 200-pound steel block, and measuring pressure produced in the temporal region of the cranial cavity with a blow to the forehead, the pressure associated with the production of a linear fracture was obtained. Since the glass being tested was also a flat, hard surface, the same type of measurement could be obtained with a cadaver head striking the glass. Tests were run at speeds as high as 35 miles per hour, and the severity of the impact to the head was determined from the measurement of the pressure produced and its time duration[8]. Thus a theory for the production of injury

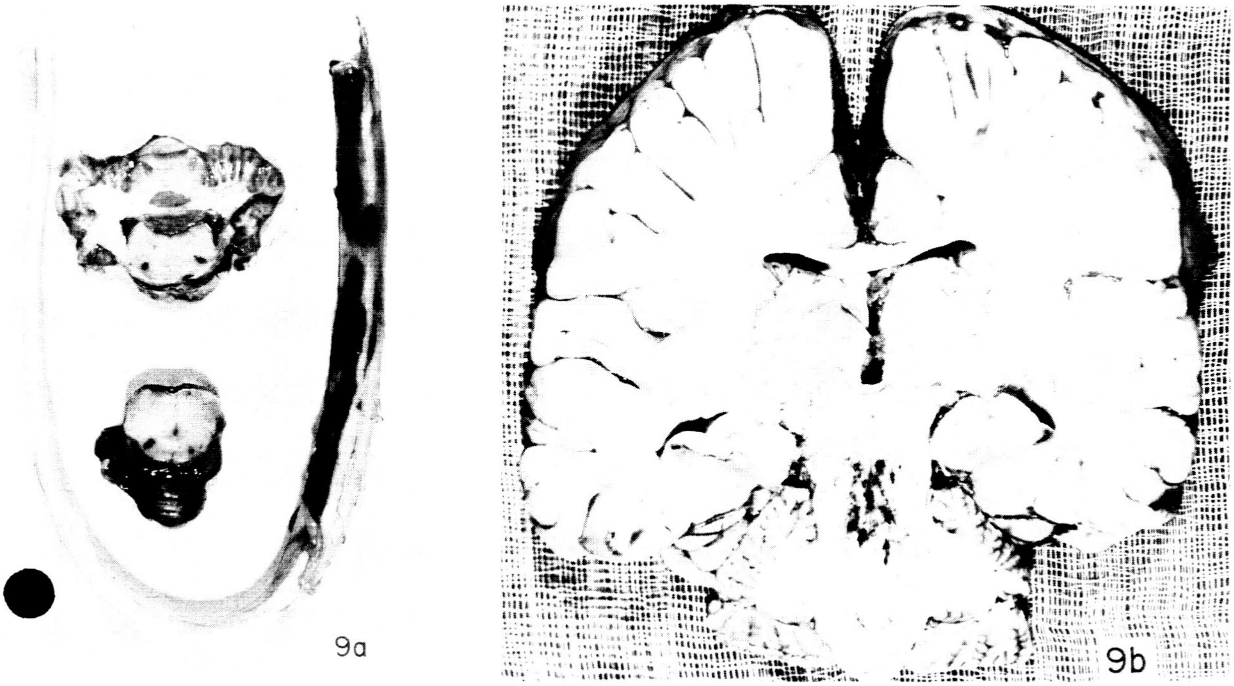


Figure 9a, b. Comparable results in the dog under anesthesia shown in "a," and in the human shown in "b," following closed-head injury. In both, hemorrhages in the brain stem are seen. In "a," the spinal cord of the animal with some subarachnoid blood is also shown.

proved to be of value; that is, if a theory can be used to explain observed phenomena, or if it can be used to predict results, it is a valuable tool in the scientific field.

Future Research

Development of theories that are useful in other areas of impact on the human body are objectives for which we should strive in future research.

Much more work is required in head impact to evaluate and refine theories which have been proposed. A comprehensive testing program should be undertaken to evaluate the mechanism of whiplash injury and to determine necessary protective measures against it. The problem of chest impact, the absorption of energy by the chest, and the damage produced by chest impact, must be studied with animal and cadaver experimentation. A study of the surges of pressure occurring in the venous system extending toward the brain due to an impact on the chest should be made and carefully evaluated. Studies of abdominal impacts causing rupture of the diaphragm and other internal damage should be undertaken to determine the conditions which produce the damage that occurs with these impacts.

Studies of long bones may be very fruitful. Impacts of the knee should be studied to determine the effects of the absorption of energy by the femur, fractures of the knee cap and the neck of the femur, and the possibility of fracture of the pelvis due to blows at the knee. Fractures of the long bones in the upper and lower extremities by transverse striking should be correlated with impact conditions which produce them. This

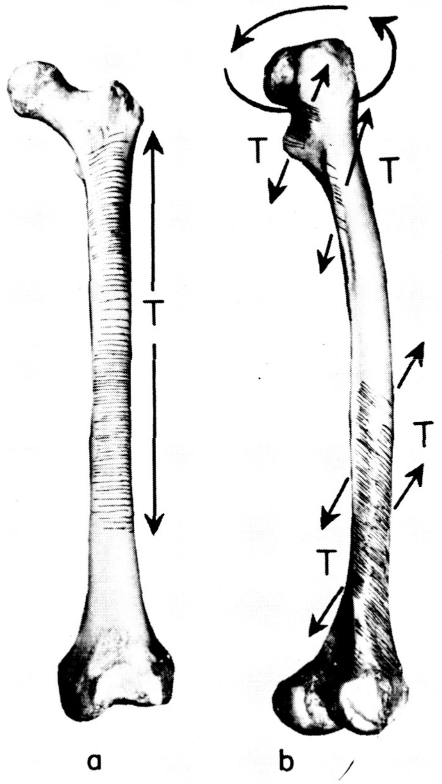


Figure 10a, b. In "a," tensile-strain pattern produced by the application of 11.8-inch pounds of energy to the middle of the posterior aspect of the shaft of the left femur of a white male, 58 years of age. In "b," tensile-strain patterns produced in the right femur of a 61-year-old white male by the application of 406.5 inch-pounds of torque(9).

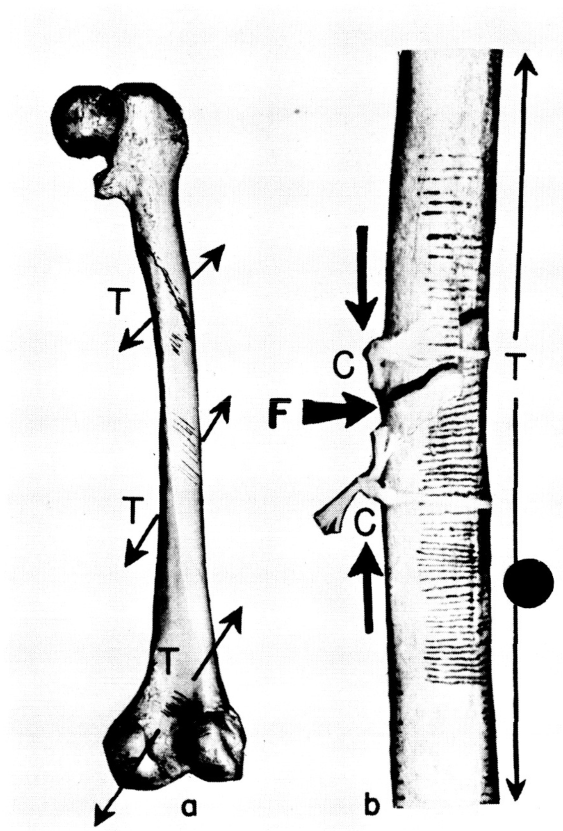


Figure 11. A spiral fracture of the shaft of the femur after application of 282.2 inch-pounds of torque. In "b," a transverse fracture of the shaft of the femur produced by a load of 390 pounds applied to the middle of the anterior aspect of the shaft. The associated stresscoat patterns may be seen accompanying these fractures(9).

list could be continued indefinitely. The relationship between a blow and the resulting injury, extent, and severity of the injury is applicable to every portion of the body.

Other investigations on clinical levels include the study of minimally, moderately, and severely injured patients by (1) electroencephalography; (2) hypothermia and its value in the management of these patients; (3) changes in the chemical and metabolic state of the body following injury; (4) study of primary shock and its relationship to secondary shock.

Electroencephalographic changes following minimal, moderate, and severe injuries to the head may give invaluable clues in management and correlation with experimental data. The effects of photic stimulation in the various classes of injured patients may give interesting information concerning brain-stem dysfunction.

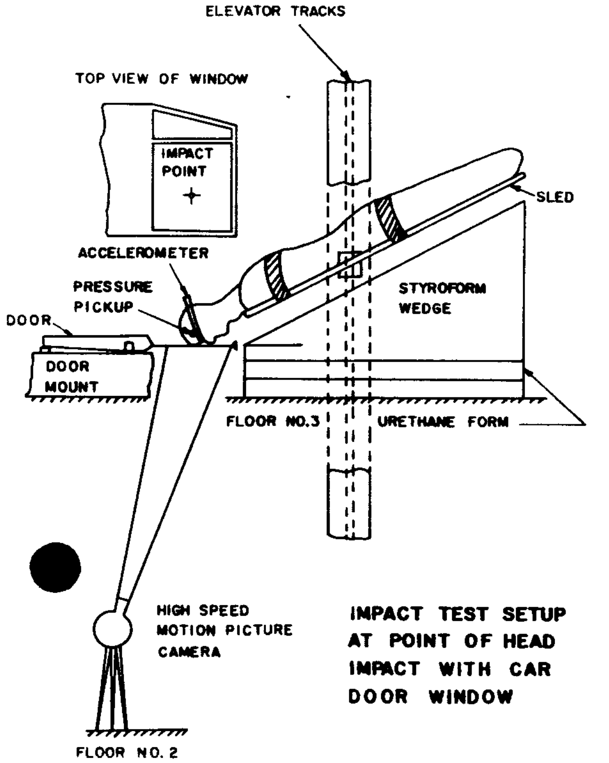


Figure 12. Experimental set-up for study of impact of head against glass. Two thicknesses and sizes of tempered glass and one of laminated glass were used in these experiments. Intracranial pressure in pounds per square inch acceleration records were obtained.

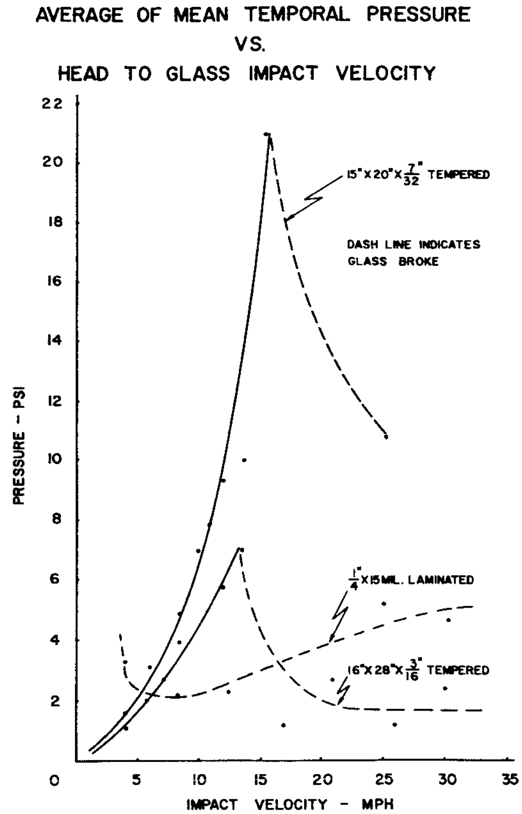


Figure 13. The findings in a group of experiments with tempered and laminated glass are shown. With the thicker tempered glass of smaller size up to 20 psi pressure was built up before the glass broke. With the thinner and larger tempered glass, break occurred at between six and eight psi. With 1/4-inch laminated glass, pressure build-up was minimal throughout and the glass broke at impact velocities of two to three psi.

The value of hypothermia in the seriously injured patient is fairly well established. Careful study of such patients by electroencephalography, and other chemical and metabolic studies for purposes of correlation, may be very worth while.

Study of the chemical and metabolic state of the patient is a large field which has been tapped only on the surface. Changes in the metabolism of sugar, proteins, and fats should be carefully studied in the human as well as in the experimental animal. Electrolytes, their levels in the blood, and their determination in the urine and sweat, may give information of value. The information may be worth while from a diagnostic as well as a prognostic standpoint.

The problem of primary shock is important in connection with brain-stem damage. In the experimental animal, we found that, following an impact causing a concussion, there was an increase in blood pressure which resulted from peripheral vasoconstriction. We found that this was mediated through the spinal cord and the sympathetic nervous system, and was not a result of adrenal-cortical liberation. We were able to show that, after adrenalectomy, increases in blood pressure occurred in the same way as when the adrenals were intact. However, the blood pressure in this type of experiment did not increase on impact when the spinal cord was sectioned in the upper cervical region. This indicated that the peripheral vasoconstriction following head impact in animals was due to a sympathetic mediation. In the human, a shock-like state with peripheral vasoconstriction, pallor, sweaty and pale skin, but with no drop in blood pressure, is frequently seen after head injury. This particular aspect of the problem, along with secondary shock from blood loss, should be carefully evaluated in the human, and this certainly can be done very effectively in certain centers in this country. One of these could well be the Detroit Receiving Hospital and its facilities.

Summary

The mechanism of skull fracture and concussion has been outlined. Strength characteristics of various bones in the body have been studied. Effects of embalming on the strength characteristics of bone have been evaluated. Impact tests upon the femur, pelvis, and spine have been conducted, and the results summarized. Recent studies on cadaver-head impacts into safety glass are briefly analyzed. Proposals for future studies include the careful study of the problem of whiplash injury, clinical studies of minimally, moderately, and severly injured humans by electroencephalography, study of chemical and metabolic changes following head injury in the human, and analysis of primary shock following human impacts.

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IMPACT DAMAGE TO INTERNAL ORGANS*

F. Gaynor Evans
Department of Anatomy
University of Michigan
and
Lawrence M. Patrick
Department of Engineering Mechanics
Wayne State University

Introduction

The present discussion will be limited to impact damage to thoracic, abdominal, and pelvic viscera; brain damage will be discussed by another participant in this conference.

Injuries to internal organs consist of contusions, lacerations, and ruptures arising from the various types of stresses and strains produced by impacts to different parts of the human body. One of the most frequent causes of injury is automobile accidents during which the occupant is injured by impact with some part of the car or by ejection from the car. In the latter case injury is produced by striking a solid object or being run over by his own or another vehicle. The magnitude of the problem is emphasized when one realizes that in 1957 there were 38,700 people killed and 2,525,000 persons injured in traffic accidents in this country.

Although more than one organ is often injured in automobile accidents, the present discussion, for purposes of convenience, will deal with the subject regionally.

Thorax

According to the annual ACIR report for April 1, 1956, to March 31, 1957, injuries of the thorax and thoracic spine were third in frequency, 26.6 per cent of 1,678 cases. Furthermore, thoracic injuries were second most common of dangerous and fatal injuries, occurring in 4.2 per cent of all injured persons. It is significant that thoracic injuries caused more deaths than did head injuries and Daughtry, cited by Kulowski[2], states that approximately 25 per cent of deaths from automobile accidents result from thoracic injuries.

The common thoracic injuries are multiple rib fractures which may be complicated by paradoxical breathing, hemothorax, pneumothorax, laceration of the lung, contusion of the heart, and on rare occasions, injuries to the great vessels, esophagus, thoracic duct, and diaphragm. Maynard et al.[3] have pointed out that normal function of the lungs is dependent upon the anatomical integrity of the chest wall, including the pleura. When this integrity is interrupted, injury of the bronchi or other air-containing elements of the lung with the attached visceral pleura may result in pulmonary

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interstitial emphysema. This can give rise to various conditions and mediastinal emphysema, regardless of its cause, is a grave complication for two reasons: (1) because of the production of mediastinal pressure which prevents adequate filling of the great veins of the heart and is rapidly fatal, and (2) because of the frequent accompaniment of mediastinal infection which, by pressure and sepsis, rapidly produces death.

Collapse of one lung from a pneumothorax is generally compensated for by the other lung, but a dangerous situation can arise when there is a progressive collapse of a lung and build-up of tension in the thoracic cavity (Figure 1). This occurs when the

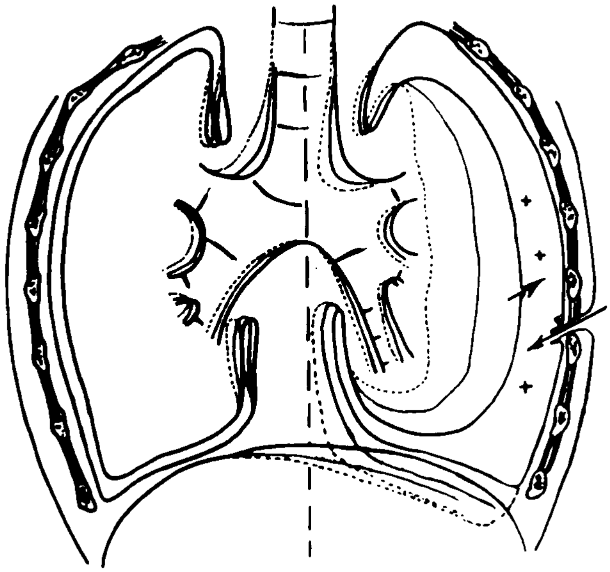

Figure 1. Diagram representing various stages in the development of a tension pneumothorax. Repetitive intake of air through a defect in the chest wall during inspiration, and failure to release it during respiration, causes a progressive increase in the intrathoracic air pressure which produces a progressive collapse of the lung, a shift in the mediastinal structures, and a depression of the diaphragm in the affected side. (From Maynard, et al., Amer. J. Surg., 90:458-468, 1955.)

visceral pleura or defect in the chest wall acts as a valve so that air can enter the pleural cavity during one phase of respiration but not leave it during the other, successive increments of air building up the intrapleural tension. This repetitious build-up of pressure completely collapses the lung on the affected side, displaces the mediastinal organs to the opposite side, angulates or tends to collapse the great veins and interferes with the filling of the heart. The result of these displacements is that the capacity of the remaining lung is lessened and the condition results fatally unless corrected promptly.

Another condition resulting from a crushing impact which destroys the semi-rigid character of the thorax is the flail-like paradoxic motion of the fractured portion of the chest (Figure 2). Under these conditions, which usually result from a fracture of adjacent ribs in two or more places, the fractured segment passively sinks in with inspiration and out with expiration. The efficiency of respiratory movements are thus decreased, and if a large area of the chest wall is involved or both sides of the thorax and the sternum crushed, the paradoxic movements must be stopped promptly or death will result.

A blow or force applied to the anterior body wall, either in the thoracic region or immediately below it, can also cause torsion of the lung, which results in gangrene because of interference with the blood supply of the lungs.

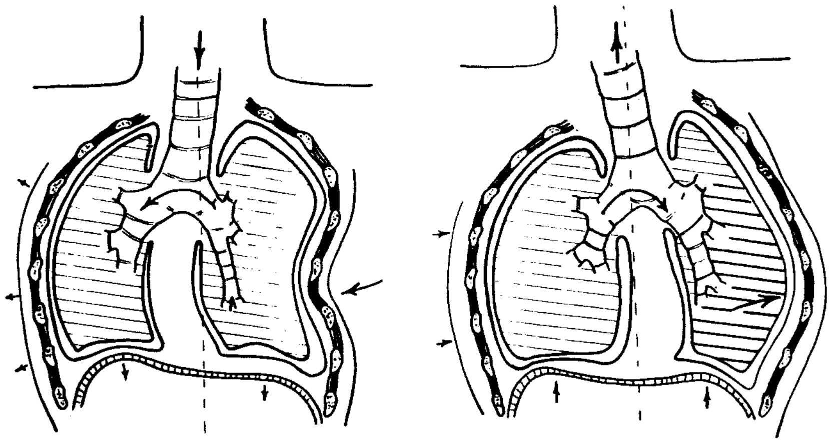

Figure 2. Diagrams showing the mechanics of flail chest. Left diagram—inspiratory phase showing the sinking-in of the unstable thoracic wall, admixture of inspired air between the lung on the damaged and the undamaged side, and shifting of the mediastinum towards the undamaged side. Right diagram—expiratory phase showing the pushing out of the unstable part of the thoracic wall, partial expiration of air from the undamaged into the damaged lung, and shifting of the mediastinum toward the damaged side. These phenomena constitute (1) paradoxic movements of the chest wall, (2) pendelluft respiration, and (3) mediastinal flutter. (From Maynard, et al., Am. Amer. J. Surg. 90:458-468, 1955.)

One rather common thoracic injury resulting from impact to the chest or upper part of the anterior wall of the abdomen is traumatic rupture of the diaphragm (Figure 3). Desforges et al.[1] reported that rupture of the diaphragm was most likely to

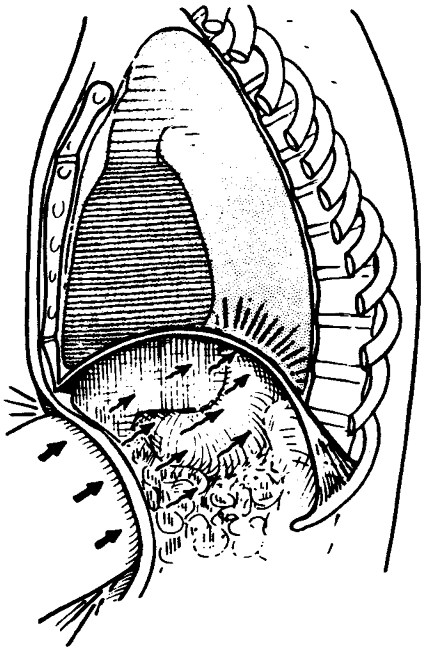

Figure 3. Diagram showing how impacts to the abdominal wall can produce forces which can rupture the diaphragm and injure the heart. (From Desforges, et al., Thoracic Surg. 34:779-799, 1957.)

occur in patients having multiple thoracic injuries. They cite that from 1,678 injured occupants of a thousand automobile crashes included in a Cornell study, 65 per cent sustained injuries in two or more body areas. All of their 16 cases of ruptured diaphragm seemed to have suffered from some form of severe impact trauma. Most of the diaphragmatic ruptures were secondary to blunt trauma resulting from automobile accidents or falls.

From the series of Desforges, et al., it appears that any type of blunt thoracic or abdominal trauma can produce diaphragmatic rupture, the relatively weak diaphragm yielding between the more solid abdominal viscera and the more pliant thoracic structures. Diaphragmatic rupture appears to be a tensile failure resulting from forces applied in an anteroposterior or bilateral direction (Figure 4). Direction of the force per se does not seem important, although diaphragmatic ruptures are infrequently associated with forces applied to the lumbar region. Abdominal viscera can, of course, herniate through the rupture so as to compress mediastinal structures and the lung.

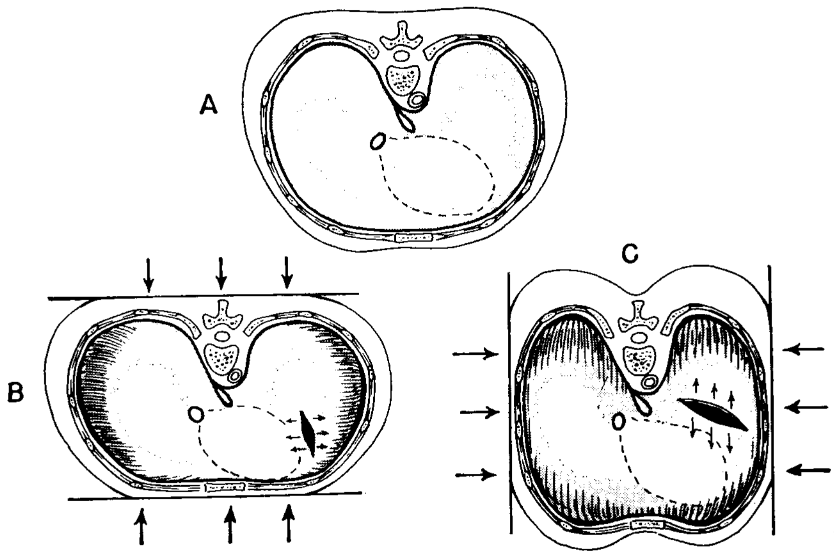

Figure 4. Diagrams showing tensile ruptures of the diaphragm arising from anteroposterior or bilateral compression of the thorax. (From Desforges, et al., J. Thoracic Surg. 34:779-799, 1957.)

Injury to the heart and pericardium may also result from an impact to the anterior wall of the thorax in the region of the heart. The mechanism believed to be involved is a secondary recoil of the heart as a result of the application of a blunt force to the external body wall. If the ribs or sternum are fractured, the recoil effect is less severe. The results of the recoil may be pericardial, epicardial, and myocardial lesions. The chorda tendineae and the heart valves may be ruptured by the sudden compressive force acting on the abdomen and the hydraulic effects of the contained blood which is forced backward toward the heart. Explosive rupture of the large vessels, the auricles, and in rare cases the ventricles, may also result.

Sudden velocity changes, as produced by steering-wheel impact to the chest, can also cause rupture of thoracic structures as the result of the action of hydraulic forces. Thus, according to Murdock, cited by Kulowski(2), the forces responsible for closed injury to the aorta are (1) inertia or "drag" resulting from sudden deceleration, and (2) compression with momentary disruption of an open circulation. The inertia force, exerted on the blood column either directly or in a torsional plane, produces frictional forces which cause wrinkles, tears, or cracks in the aortic intima. If a simple direct weight is exerted strongly enough against a fixed point, complete transsection of the aorta may result from shearing forces. Momentary stoppage of cardiac flow by compression may also produce an effect on the aortic wall similar to that produced by squeezing a full-blown balloon.

Kulowski(2) also cites some investigations of Rice on the mechanics of closed aortic injury. Thus, impaction of the chest against a steering wheel causes the central part of the descending aorta, which is less rigidly fixed, to be snapped forward by the decelerative force and the mass of the contained blood. The forces involved are absorbed by the elasticity of the aorta and the thoracic viscera. However, the rate of deceleration is not the same for all parts of the aorta. Thus, the upper part of the aorta, which is more rigidly fixed by the great vessels arising from the arch and the ductus arteriosus, decelerates at the same rate as the body while the less rigidly fixed descending aorta decelerates at a different rate. The force resulting from these differences in the rate of deceleration is concentrated directly below the ligamentum arteriosum, the point of maximum fixation and the classical site for the rupture of a healthy aorta as a result of indirect violence.

It was noted by Hass, cited by Kulowski(2), that in 1,740 persons involved in 118 aircraft accidents intrathoracic injuries of the heart, lungs, aorta, and serous surfaces were particularly severe when accompanied by fractures of the thoracic wall or vertebrae, although they were also present in the absence of fractures. The limiting factor appeared to be the elastic limit of the viscera to stress rather than tolerance of the bones to fracture. Kulowski cites some experiments made at Ohio State University in 1952 which showed that isolated ribs were distorted an average of 22.6 per cent before fracturing. In tests with intact cadavers, in which the load was applied by means of a four-inch-wide belt around the thorax, loads of more than 5,500 lbs. fractured only a single rib. The ribs of a deviscerated cadaver could not be fractured, although the sternum could be pushed back against the vertebral column.

Trauma of the heart, pericardium, and great vessels results most frequently from frontal blows to the thoracic cage, although they may also be produced by prolonged pressure which squeezes the heart between the sternum and the vertebral column. The heart and great vessesl can also be injured by abdominal viscera which are pushed up against the diaphragm by blows to the abdominal wall. Impacts and crushing forces applied to the lower limbs may initiate shock waves in the blood stream which may be reflected against the heart walls and injure the myocardium by stretching it beyond its elastic limit.

Abdomen

The abdomen is less frequently injured than the thorax. Thus, in 800 motorist crash hospital patients reported on by Kulowski(2), injuries to the thorax and back occurred in 36.87 per cent, the abdomen and lumbar region in 17.82 per cent and the pelvis in 12.37 per cent of the cases. A summary, taken from the same author, of the frequency of injury to various organs in 130 motor-vehicle-accident fatalities is given in Table I.

TABLE I

Frequency of Primary Injury to Body Organs Among 130 Motor-Vehicle-Accident Fatalities (Data from Kulowski, '60)

Organ	% Injury	Organ	% Injury	Organ	% Injury
Lungs	7.2	Stomach	1.2	Mesentery	1.4
Pleura	3.2	Liver	3.8	Kidney	1.6
Heart	5.4	Spleen	3.4	Suprarenal gland	1.2
Pericardium	1.2	Pancreas	0.6		
Great vessels	1.2	G. I. tract	2.4	Urinary bladder	3.2
Diaphragm	0.6			Ano-rectal	0.8

The abdominal structures most commonly injured in automobile accidents may be classified into three groups: (1) solid organs such as the liver, spleen, kidneys, supra-adrenal glands and pancreas; (2) hollow organs such as the stomach, small and large intestines, urinary bladder, and gall bladder; and (3) supporting structures such as mesenteries, peritoneal reflections, blood vessels, and nerves. Trauma to the supporting structures and to the solid organs may produce lacerations and tears resulting in hemorrhages. The hollow organs vary in size according to the time of day and their content; thus when they are distended with fluid, the tendency to rupture is greater than when they are empty.

According to Moseley and Zeller's[5] investigation of results of deceleration forces on 1,740 occupants involved in 118 aircraft accidents, lesions of the abdominal wall were generally unimportant. However, intra-abdominal lesions were common and consisted of lacerations of the liver, spleen, ilium, and mesenteries. Hass, cited by Kulowski[2], classified lesions into four types: (1) external injuries due to point-to-point contact between the surface of the body and an external restraining structure, (2) injuries resulting from deformation of deep-seated tissues or viscera adjacent to the points of the body contact on the external surface, (3) injuries arising from the action of forces on tissue structures, independent of direct contact, and (4) injuries developed after the force had been applied. Examples of various types of hepatic injury produced, according to Moritz[4], by blunt impacts to the anterior abdominal wall are illustrated in Figure 5.

The relation of forces to the internal injuries was studied experimentally in mice by Rushmer[6]. The high incidence of interstitial tissues within the lungs of the experimental animals suggested that the impact produced pressure waves which were transmitted through the body tissues and fluids causing rupture of the relatively unsupported blood vessels of the lungs. The relatively high incidence of lacerations of the liver and spleen suggested that these organs were placed under stress due to distortion as a result of the sudden displacement of the abdominal contents. Because the liver and spleen are not as flexible as other abdominal organs, they are more apt to develop lacerations at or near the surface in response to sudden changes in their shape. In the supine position, the forces acting on the abdominal surface of the mice used in experiments were calculated to range from 157 to 227 g.

The internal injuries produced in cats, rabbits, and mice were compared to those found in fatal crash injuries in man by Rushmer and Hass[7]. In more than 75 per cent of the cases in both man and the experimental animals, gross pathological

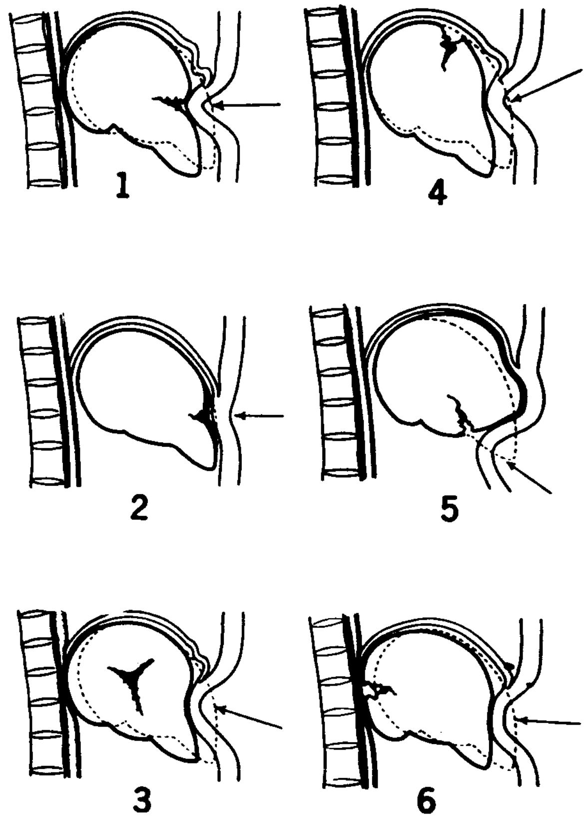

Figure 5. Diagrams showing common types of hepatic lesions resulting from blunt impacts to the anterior abdominal wall. (1) Transcapsular laceration immediately beneath the side of external impact. (2) Subcapsular laceration beneath the site of external impact. (3) Non-communicating central laceration. (4) Coronal laceration due to distortion. (5) Laceration of inferior hepatic surface due to distortion. (6) Contrecoup injury. (From Moritz, The Pathology of Trauma, Lea and Febiger, 1954.)

changes were found in the lungs. The type of lesion seen in the lungs, liver, spleen, diaphragm, kidneys, pancreas, and G. I. tract in experimental animals were similar to those in man. However, the incidence and severity of the lesions were greater in the humans who died instantaneously.

The damage or injury produced in the human body by acceleration is because the internal organs behave as visco-elastic materials. Furthermore, the magnitude of the stress and acceleration, or dynamic response, can be increased by the elasticity of the human torso, as is known from the theory of elastic structures.

Proposed Experimental Program

With some notable exceptions, experimental investigations of trauma of the internal organs and attachments due to impact have not been as numerous nor the results as definitive as those relating to the skeletal parts. Pathological and clinical studies of damage to the internal organs from accidents have been rather extensive, but give results based on speculation concerning the mechanism causing the injury. Comprehensive programs are needed to:

a. Establish criteria of degree of injury to the various organs.
b. Develop instruments and techniques for measuring the degree of injury.
c. Determine the mechanism causing the injury.
d. Devise means for minimizing injury from impact.

In general, the experimental program can be divided into three broad categories:

a. Human experimentation in which the impacts must be kept at a sub-injury level, with tolerance limits established, and perhaps some degree of injury information inferred by extrapolation.

b. Animal experimentation wherein all degrees of injury up to and including complete destruction of the organ are attainable, but the correlation with humans is an unknown factor.

c. Cadaver experimentation with the problem of interpreting the results from dead to living organs.

While all three approaches have major drawbacks, a judicious analysis of the results of all three, together with the clinical and pathological studies, should permit an interpolation between tolerable and fatal injuries with good accuracy. Voluntary tolerance limits form the low injury limit, while fatal injury or complete destruction of the organ (from pathological studies) provides the other endpoint for living human organs. These endpoints can serve as guides in both cadaver and animal work.

Similar endpoints can be established in living animals for the various types of impact. Intermediate injury as a function of the important impact parameter(s) will then be determined, and the relationship applied with suitable modifications to human organs. This sounds like a straightforward experimental procedure, but it will be complicated by physiological variation from one animal to the next, difference in animal and human organs, difficulty in evaluating the effect on the organ, and problems associated with sensing and recording the injury symptoms reliably.

Identification of which, if any, of the organs can be studied from cadavers must be given first priority. The effect of impact or acceleration on the function of the organ cannot, of course, be determined. It may be possible to establish a relationship between functionally and some measurable quantity such as pressure displacement or distortion. Even if the organs from cadavers can't be tested as such, the load distribution to the organs can probably be ascertained from cadavers.

Types of Injuries to be Investigated

The several types of injuries to be investigated can be subdivided into the following parts for convenience.

a. Lacerations or lesions due to impact on sharp or blunt objects.

b. Lacerations or lesions from inertial forces. Damage to organs or organ attachments can be found as a result of these inertial forces.

c. Hydraulic damage which can cause rupture of organs from internal pressure due to force applied directly to the organ, and damage to an organ from hydraulic pressure at a remote point.

d. Lacerations, lesions, or crushing from impact or restraint over large areas of the chest or abdomen. This type of injury corresponds to that found in automobile accidents where a steering wheel or large areas of the dash panel contact the chest or abdomen.

Many injuries will fall into more than one of the four categories outlined above. It will be difficult to isolate the causes of multiple injuries, and in fact there will be many cases in which they cannot be separated, since several types of injury may result from a single impact.

Investigation of injury to the internal organs is complicated by the many organs involved. These include: kidneys, spleen, adrenals, colon, pancreas, stomach, small bowel, large bowel, bladder, gall bladder, lung, heart, aorta, esophagus, thoracic duct, diaphragm, blood vessels, nerves, mesentery, and fassia. Some of these can be eliminated, since injury to them is rare and often associated with major injury to other organs which overshadows the less serious injury. Clinicians and pathologists will have to determine which of the organs are most important in relation to number and severity of injuries, and their overall effects on normal human functions. One of the major contributions of a detailed study of individual organs and the effects of impact on them will be a better understanding of which organs are most likely to be damaged by particular impacts, thereby aiding the physician in diagnosing injuries after accidents. Injury to particular organs is often not immediately evident, and the patient is treated for an obvious but not so important injury, while the main injury, which can cause permanent damage, is left untreated until complications arise.

Investigative Techniques

Of the three broad experimental categories described heretofore, the one that probably will provide the most information is that in which living animals are tested. A proposed procedure is as follows:

Living animals will be subjected to impacts of the various types already described, with the effect on particular organs noted. The same animals or other animals will then be sacrificed, subjected to the same type of impact, and the same measurement(s) noted on the dead animals. A correlation between living and dead animals will then be available. The same or similar correlations will be applied to humans. A further approach is to attempt to duplicate the pathological injuries noted from accidents on cadavers. There are many difficulties, of course, in using cadavers, due to the changes in organ tissues, different hydraulic reactions, and lack of physiological response. However, efforts will be made to use cadavers as realistically as possible. For example, certain of the organs can probably be injected with a suitable fluid, with restrictions at all entrance and exits to the organ. It should then react somewhat similarly to living organs with respect to hydraulic pressures or direct impacts.

Several different methods of testing have already been utilized to some degree, and others are proposed. In the first place, the most obvious impact test is to drop a known weight of a given shape on the organ in place in the cadaver or animal, or, conversely, to drop the entire cadaver or animal on the impact form. Both techniques have been used at Wayne State University. Some of these results have been or will be described by other investigators at this meeting. The equipment available for this type of test should be a drop tower of adequate height to achieve the desired velocity. Generally this height does not have to be excessive and is available in most multistory buildings with a free area from one floor to the next. Guide rails or wires are desirable to ensure correct orientation at instant of impact, and a means of raising the body or anvil to the predetermined height with a remote-controlled, quick-release mechanism is a convenience.

In addition to the drop tests, a different type of "impact" is obtainable through controlled acceleration. The difference between impact tests and acceleration tests is simply a matter of degree, since in both cases it is actually the accelerations with accompanying inertial forces that normally cause damage. Impact accelerations are generally of much higher magnitude and shorter duration than are found in most linear accelerators or centrifuges. There is a vertical linear accelerator at Wayne, which will also be described by other investigators at this meeting, that provides controlled accelerations of any desired value from 0 to 50 g's, with time durations from 0 to one second or longer, depending upon the acceleration magnitude. The rate-of-onset or shape of the acceleration pulse is also important, so the test equipment should have provision for controlling these.

A horizontal linear accelerator is essential, since it permits horizontal impacts with the body in standing or seated attitudes similar to those of pedestrians or passengers in automobile accidents. A horizontal accelerator should have a controllable and variable rate-of-onset and acceleration-magnitude. An impactor such as a rotary hammer or linear actuated hammer is also an important tool, since it permits impact over small areas with a high degree of accuracy. A pressure gun or valve, allowing controlled pressures to be applied directly to the organs, will permit study of the hydraulic effects on individual organs. Such an instrument is used in brain-injury studies at Wayne State University. The pressure-magnitude and duration of application is controlled. Another useful pressure device is a tank with controllable pressure pulses. The tank should be large enough to permit an entire cadaver or animal to be immersed with pressures ranging from those encountered by divers to those corresponding to explosions.

Instrumentation

One of the major requirements is that instrumentation must not interfere with or modify the response of organs. Thus, any sensing element that is inserted into an organ must be small enough so that no damage occurs during insertion and no functional change results from its presence. Where possible, a sensing element should be inserted through the normal openings of the body or at least through the normal openings of the organ under test. Such transducers or sensing devices are generally not available for purchase. Consequently it will be necessary to develop special transducers for use in this work.

Pressure transducers that can be attached to the end of catheters and are no larger than catheters are required for insertion through arteries, etc. Such a transducer is shown in Figure 6. It was developed for a study of pressures in the cranial cavity near the foramen magnum. Several other pressure-sensing devices are included in the same figure. They were developed for particular uses such as measuring pressures just inside the skull, and, in the case of the "Stem" pick-up, to measure pressure-distribution throughout the skull.

Accelerometers are essential, since they indicate the actual acceleration experienced by the organ, which is very often considerably different in magnitude from the applied acceleration and may be completely out of phase with the applied acceleration. Many small accelerometers with the required sensitivity and frequency response are available. The crystal type is usually the smallest with the highest-frequency response and sensitivity, but has the disadvantage of improper readings at low frequency. In impact work, the poor low-frequency response is generally not a serious drawback. Crystal accelerometers are available less than one-quarter inch in diameter by one-quarter inch long.

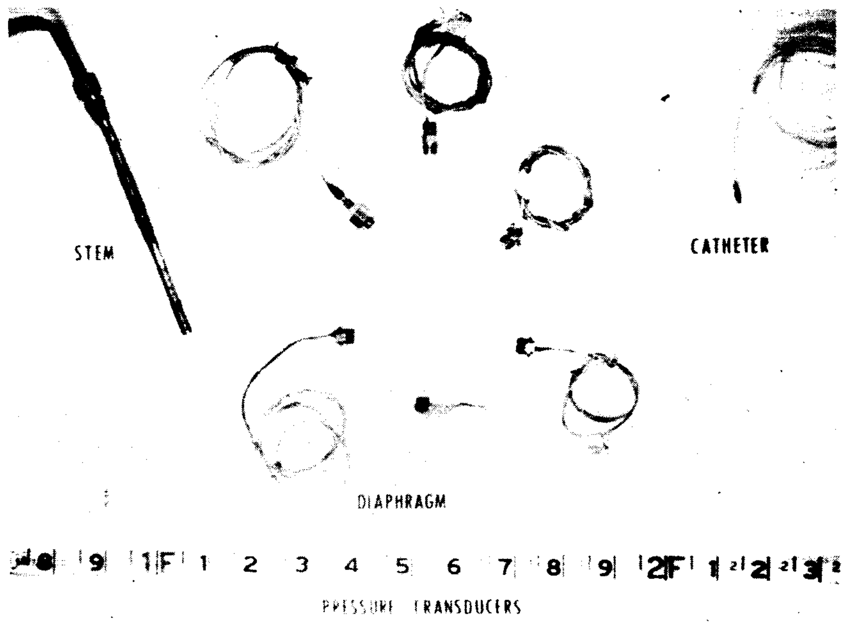

Figure 6. Photograph of some of the pressure transducers developed for use at Wayne State University.

Strain gauge, reluctance- and capacitance-type accelerometers have also been used with good success. The choice depends on the particular application.

Force-measuring equipment, to measure the force between an organ and its attachments, is a valuable adjunct to the accelerometer. Extensometers to measure distortion and stretching of tissues will provide needed information. These could take the form of free or unbonded strain gauges which are actually sewn into the tissue and will react with the tissue. Temperature-sensing devices including thermo-couples and thermisters are available, but will have to be designed into a package suitable for use in the organ.

An example of the special transducers that must be designed for this work is shown in Figure 7. The tendon dynamometer was initially made to measure in vivo force in the human achilles tendon. This is accomplished by deflecting the tendon slightly as shown in the diagram, and measuring the comparatively small transverse force resulting from the longitudinal tension in the tendon. Calibration is achieved by applying a force to the ball of the foot with the dynamometer in place, and calculating the force in the tendon from the known force and simple geometric relationships. The smaller one shown in the diagram is being used on cats. Much smaller ones can be m[illegible] using three needles to deflect small muscles, ligaments, or other organ attachments so in vivo measurements in animals can be obtained under impact.

The usual measurements of EEG, EKG, respiration-rate, external blood-pressure measurements, and so forth will of course be used to provide data on the effects on the main body functions of injury to particular organs.

High-speed photography is invaluable in permitting a visual observation of the movement and, in many cases, the progressive damage up to rupture of the organ.

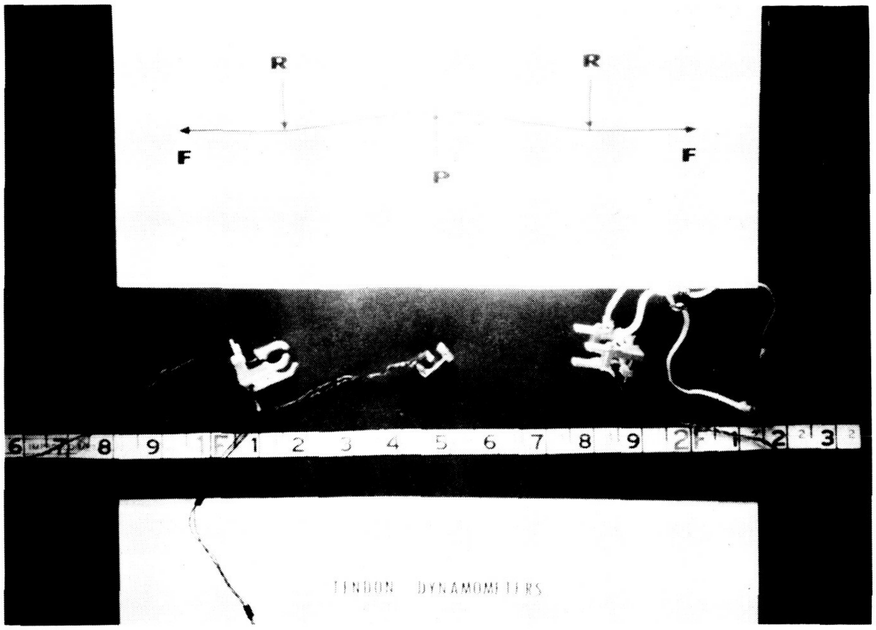

Figure 7. Photograph of tendon dynamometers and a diagram showing operating principle. Wayne State University.

Past experience has shown that rates of 5,000 to 6,000 frames per second are required to permit adequate analysis of impact, with higher rates desirable in some instances. Windows cut through the outer layers of tissue, and covered with a transparent material, permit observation with the organ in place. Good results are also being obtained with photography through tubes. Actual rates of deformation as well as total deformation and rupture can be determined from such photography. High-speed x-ray photography permits the internal viewing of the organs. This is especially true if the organs are tagged with a radio-opaque material, such as a small clip which will not change the effective mass of the organ, or by ingestion of a radio-opaque material using normal techniques. It may be possible to determine such things as total displacement by means of normal x-ray equipment. This could be accomplished by applying the impact while the x-ray equipment is on. In that case the total excursion will be indicated by the excursion of a tagged part on the x-ray film, or the organ itself may be visible in shadowy form on the x-ray. Also, the rate of travel or velocity of the organ may be roughly determined by the density of the x-ray pattern. The pattern will show a denser form at low or no velocity, and less dense at high or maximum velocity.

It is essential that all measurements be recorded so that there is no operator error in reading the output of a sensing element. Furthermore, it is generally necessary to know the time history of the information being recorded. For example, measuring pressures and the effects of pressures in the cranial cavity, it is important to know not only the peak acceleration but also the duration of the pressure and the shape of the pressure pulse. It has been found that high pressures of extremely short duration may not cause damage, while a much lower pressure applied over a longer period will cause serious damage. The recording equipment should show the results immediately if possible, so the results can be examined for adequacy. Then, if the required information is not suitable, the experiment can be repeated immediately before going on to some other experimentation. The recorder must have adequate frequency response so that the peaks and short-duration variations are recorded.

Much has been accomplished in the area of impact injury to the internal organs, but there is far more still to be accomplished. Best results will be attained through the close cooperation of physicians, anatomists, and engineers, including the many special fields represented in these groups.

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COMPARISON OF THE DYNAMIC CHARACTERISTICS OF DUMMIES, ANIMALS AND MAN

Rolf R. Coermann
Aerospace Medical Research Laboratories
Wright-Patterson Air Force Base, Ohio

Introduction

The use of animals and dummies as substitutes for the human in acceleration, impact, and vibration research is desirable to reduce the hazard for man. This would be possible if the dynamic response of the substitute were the same as that of the human body. The determination of the mechano-dynamic characteristics of the human body and the substitute, for the type of excitation to be studied is, therefore, imperative.

The conventional method of determining the dynamic response of a complex system within a certain frequency range is to vibrate the system and measure its mechanical impedance or mobility(1). For linear systems the parameters for the different resonances can be calculated from an impedance curve and a mechanical model can be derived as a simplified substitute for the complex system in the frequency range studied(2). The response of this model to another type of excitation, for instance, a single deceleration pulse, can then be calculated. This method can be applied to dummies, which are mainly linear in their response.

However, living organisms have linear characteristics only in the low-frequency range and for very small deformations. For higher transmitted forces a non-linear response must be expected, and even the magnitude of the parameters for the different resonances will vary. The damping coefficients, for instance, may depend on the instant velocity of the organ displacements, deviating from the assumed viscous damping, and the participating mass may change with higher accelerations. There will be a difference, therefore, between the response of an organism to steady-state vibration with relatively low acceleration and to impact pulses with short high accelerations. This consideration holds for the responses of the whole body as well as for a single organ or organ complex. These differences must be investigated experimentally.

When animals are used as a substitute for man, the differences in body configuration will be another problem. If the natural frequency of the studied organ system differs much from the related organ of man, their responses to a certain type of excitation cannot be compared. Also the posture of the living subject will have an influence on his dynamic response, necessitating a careful control of the subject's posture during the test. Establishing tolerance criteria for man, using animals as substitutes, presumes that the tissue strength in animals is the same as in man, which is very unlikely and not yet investigated.

Summarizing these considerations, it becomes evident that before using substitutes for man in dynamic tests, their dynamic characteristics for the type of excitation

studied must be carefully tested, and any conclusion from such a dynamic test must be related to the proper parameters of man.

Dynamic Responses of Man, Animals and Dummies to Steady-State Vibration

The average response of sitting human subjects to steady-state vibration in the frequency range 0 to 20 cps is illustrated in Figure 1. Up to two cps the impedance of

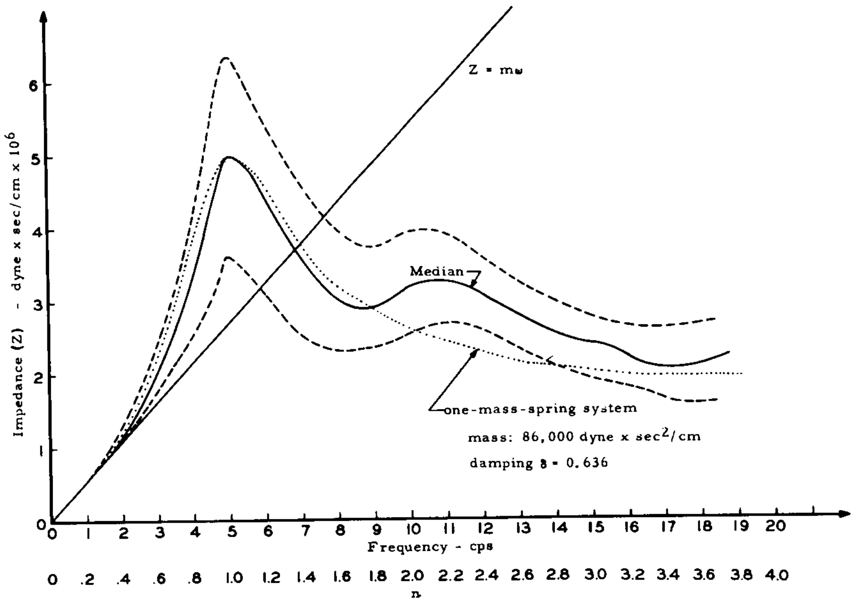

Figure 1. Median and the twentieth and eightieth percentile of the impedance of eight different subjects, compared with the impedance of a pure mass (mw) and a one-mass-spring system with damping.

the whole body is the same as the impedance of a pure mass (mw), indicating that the body moves as a whole. Around five cps a resonance peak produces the greatest deformation of the whole body, and a second peak around 10.5 cps demonstrates a resonance of a subsystem in the body. The calculation of the parameters for these two resonances leads to a model consisting of two mass spring systems with dampers. The first system (dotted line) represents about 90 per cent of the body mass, has an elasticity of about 685 lbs per inch and a damping factor of about 0.32 of the critical damping; the second represents about 10 per cent of the body mass, has an elasticity of about 186 lbs per inch and a damping factor of about 0.13 of the critical.

These parameters change considerably if the subject assumes a relaxed position or if the abdomen of the subject is restrained by a semi-rigid envelope (Figure 2). While the first resonance frequency changes only little, the damping factor for this resonance varies over a wide range, reducing the height of the peak almost to the mw-line, indicating that the body reacts almost as a pure mass for frequencies up to six cps. The second resonance practically disappears.

The same effect can be seen if the transmission of vibration from the vibratory table to the head of a sitting subject is measured (Figure 3). The peak of the fundamental resonance at around five cps is reduced in a relaxed posture, and the higher resonances become insignificant.

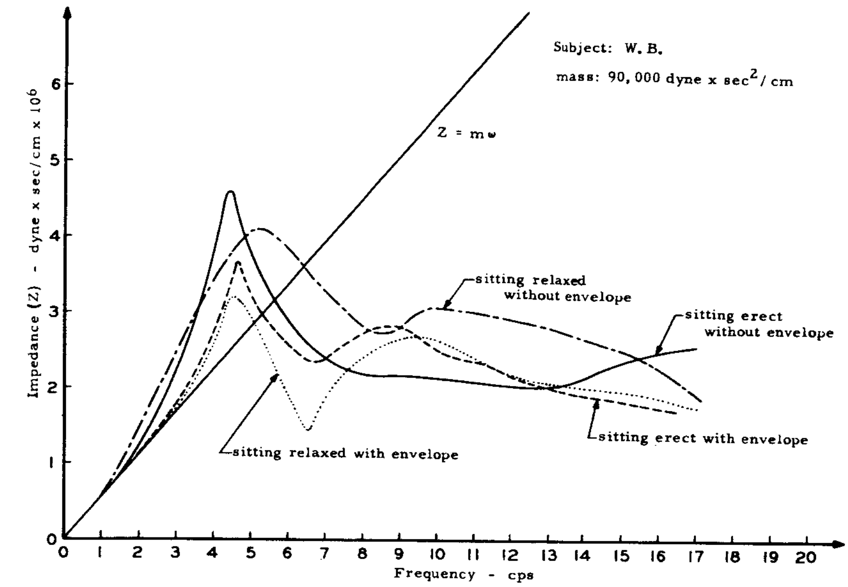

Figure 2. The effect of a semi-rigid envelope around the abdomen on the impedance of one sitting subject at erect and relaxed postures.

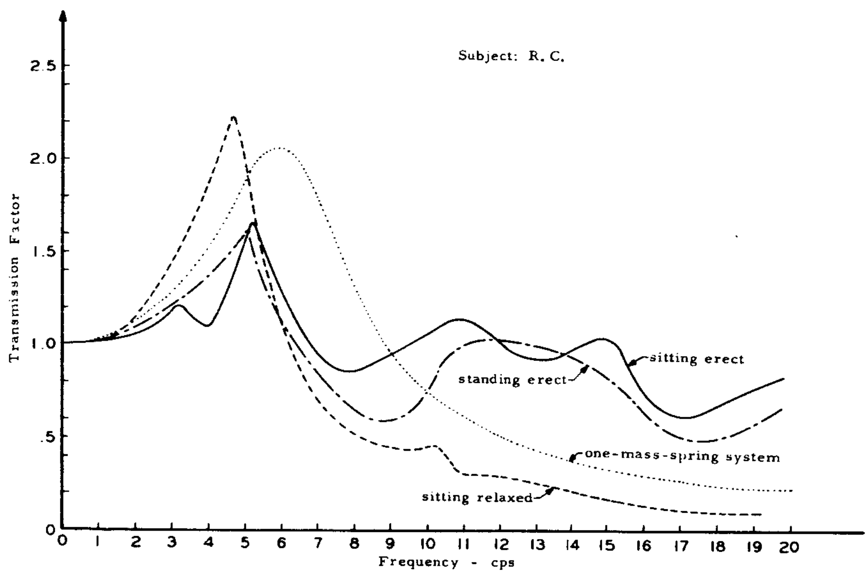

Figure 3. The transmission of vibrations from the seat to the head of one subject at varied body postures compared with the transmission factor of a one-mass-spring system with damping.

In semi-supine position, the response of the human body to steady-state sinusoidal vibration is completely different, as shown in Figure 4. The general course of the impedance curve follows very closely the mw-line up to about 15 cps, except at around eight cps where a slight peak due to a resonance in the pelvic area becomes evident. The second enhancement of the curve around 14 cps indicates a very high damping in the body, and the phase angle signifies the domination of the damping forces at higher frequencies.

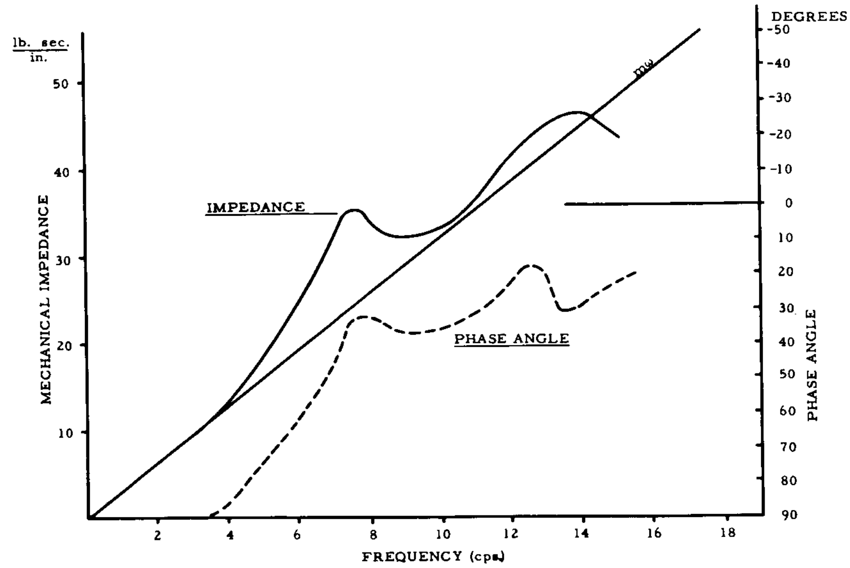

Figure 4. Mechanical impedance and phase angle of man semi-supine.

One species of animals used as a substitute for man in impact research was the bear, because it was assumed that his body structure is most similar to that of man. The steady-state impedance curve (Figure 5) of a 126-lb Himalayan bear proves that

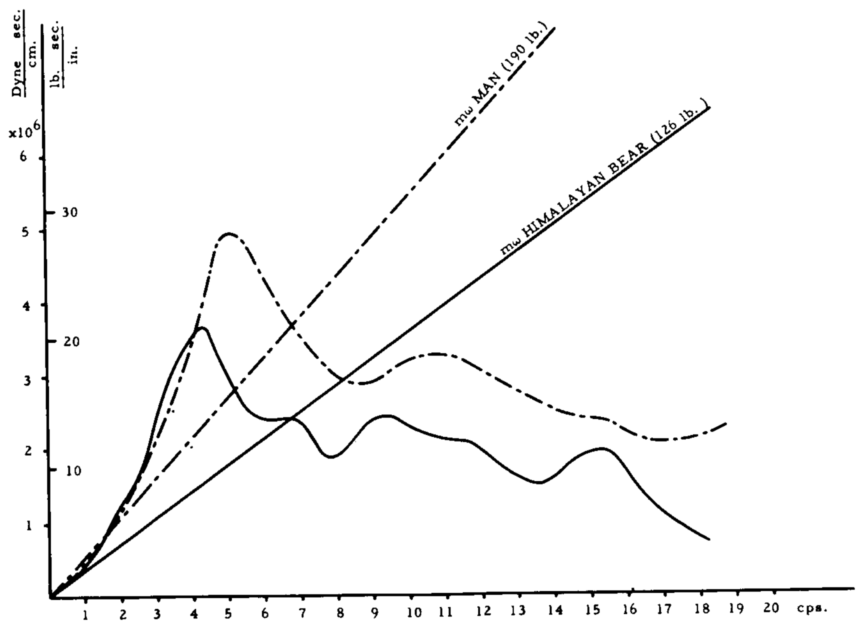

Figure 5. Impedance of a sitting Himalayan bear compared with impedance of man.

this assumption is about right. The first resonance frequency is a little lower than that of the average man, but still in the range of the standard deviation. Also at higher frequencies the characteristic of the impedance curve is similar if the difference in weight is considered.

The movements of the visceral organ can be a limiting factor for human tolerance to steady-state vibration in the frequency range four to eight cps. Also at impact, these organ displacements can produce internal injuries, especially if they are excited at their resonances. To measure the resonant frequencies of the thoraco-abdominal system as a whole, an air-filled balloon on a catheter was inserted in the colon of human and animal subjects sitting on a vertically vibratory shake table(3). The changes of the pressure within the balloon and the phase of this pressure relative to the movement of the shaker was recorded at frequencies from one to 20 cps. In Figure 6 the pressure changes per g of the shaker are plotted versus frequency for a man sitting

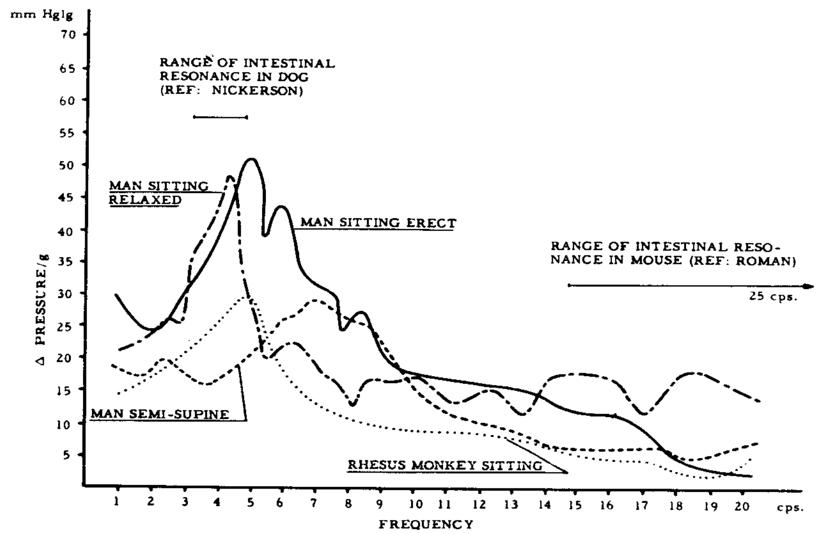

Figure 6. Comparison of intestinal pressure change during sinusoidal vibration in man at different postures and in rhesus monkeys (anesthetized).

erect, sitting relaxed, and semi-supine, as well as for a sitting rhesus monkey. Man and monkey show in the sitting position main resonances between four and five cps, while the man in semi-supine position has the maximum movements of the thoraco-abdominal viscera between seven and eight cps. Both findings coincide with the impedance measurements. Nickerson determined by an x-ray method the resonances of abdominal viscera of anesthetized dogs to be from three to five cps(4) with similar damping coefficients, but Roman(5) found that the abdominal resonances of mice measured externally lay in the frequency range between 15 and 25 cps. Thus, before animals are used to study the influence of transient forces on the abdominal organs, it must be assured, that their thoraco-abdominal systems have similar dynamic characteristics to that of man.

Three different types of dummies have been tested in this laboratory:

1. Sierra Dummy 1, weight 187 lbs
2. Sierra Dummy 2, weight 212 lbs
3. Alderson Dummy, weight 215 lbs

These dummies were restrained in a standard Air Force seat with standard harness and the mechanical impedance for vertical sinusoidal vibration was measured. In Figure 7 the impedances and phase angles of these dummies are compared with the

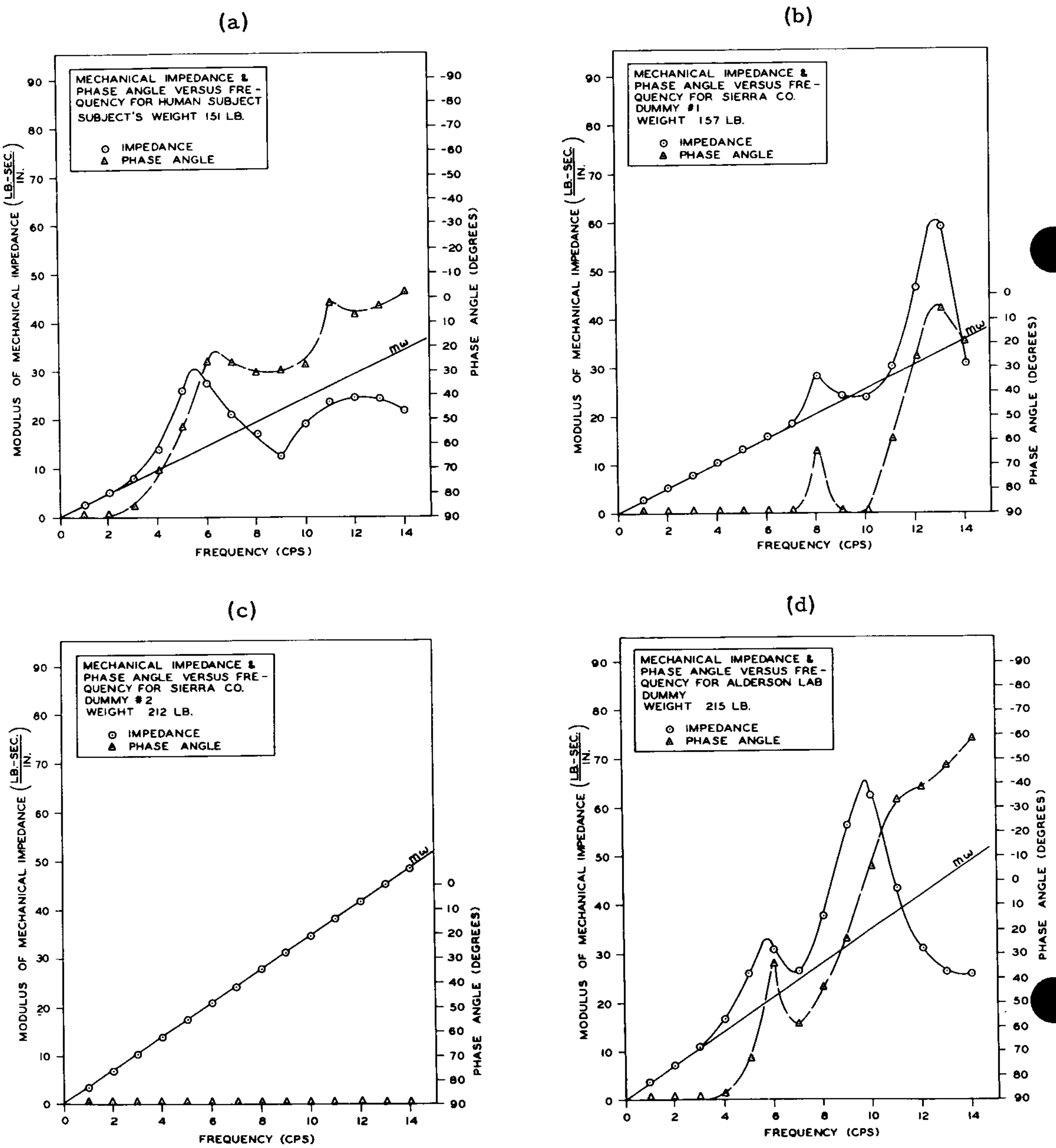

Figure 7. Mechanical impedance of man (a) and three different dummies (b, c, and d) sitting in a standard AF seat and restrained with standard harness.

impedance and phase angle of an average-sized man under equal conditions. The human body shows again the two resonances around five and 11 cps while the resonances of the dummies are:

Sierra Dummy 1: at eight cps (slight) and at 12.9 cps (dominating)
Sierra Dummy 2: no resonance (pure mass)
Alderson Dummy: at 5.7 cps (medium) and at 9.8 cps (prominent)

The differences between these four tested subjects are evident; therefore, they can not be expected to respond in the same manner to a given dynamic input. But, since the parameters of these dummies are also practically linear for high-impact loads, they are very suitable for proving the adequacy of the dynamic test method described in the following paragraph.

Dynamic Responses of Man, Animals and Dummies to Impact

Theory

The theory of transients in mechanical systems reveals the correlation between the natural frequency of a single mass-spring system and the duration of an impact pulse applied to such a system[6]. Figure 8 illustrates this relationship for a sine

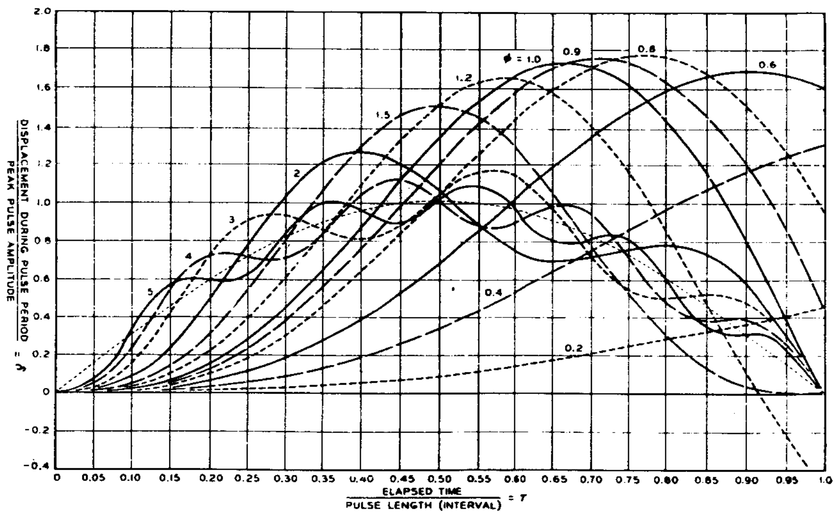

Figure 8. Transient displacements of a mass-spring system subjected to a sine whip for various values of Φ[6].

whip. The transient time displacement of the mass is plotted versus the relative elapsed time during the duration of the whip for different ratios of pulse length to natural period of the system (Φ). If the natural period of the system is short compared to the pulse length ($\Phi>2$), then the mass follows essentially the displacement of the pulse, and oscillates around this movement at its resonant frequency. If $\Phi<1$, the system becomes excited by the pulse during a relatively short time and oscillates after the pulse according to its energy content. For $\Phi = 0.8$, the mass has the maximum displacement during the pulse, which means that the transmission factor (mass movement to pulse movement) is at its maximum. If the system has damping, then the mass displacements are reduced, because the pulse energy becomes absorbed in the

damper after a short time; however, the relationship of the transmission factor to Φ remains about the same. Another revealing factor is the phase-shift between the peak of the mass movement and the peak of the pulse movement. In Figure 8 it can be seen that, without damping, the maximum mass movement for $\Phi = 0.8$ occurs at 0.775 of the pulse duration, or about 50 degrees after the pulse maximum. These two criteria, maximum transmission factor and 50 degrees phase-shift, can be used to detect the fundamental resonance of a system subjected to an impact pulse.

Instrumentation and Test Method

To investigate the response of systems to impact pulses, it is necessary to have a facility that can produce defined and highly repeatable deceleration patterns. A drop tower has been developed for this purpose at the Aerospace Medical Research Laboratories, W-PAFB, Dayton, Ohio. This device consists of a cart guided by two heavy rails and accelerated by gravity during a free fall of up to 30 ft. A plunger is attached on the cart and enters a water-filled cylinder at the end of the free fall. The acceleration of the water through the small gap between plunger and cylinder produces the deceleration of the cart. The shape of the deceleration curve depends upon the shape of the plunger, the drop height, and the weight of the loaded cart. The test subject is mounted on a stiff platform bolted on the cart. Human and animal subjects as well as the tested dummies were restrained in the sitting position on a rigid chair mounted through three force transducers on this platform. All three force transducers were connected together electrically and had the same sensitivity, so that their combined output was independent of the location of the subject's center of gravity. Records were taken of the deceleration of the seat and the force transmitted to the seat. For evaluation of the record, the inertia of seat (deceleration times mass) was deducted from the transmitted force at any instant of the pulse and plotted together with the seat deceleration versus the elapsed time (see Figure 9). The transmission factor was established

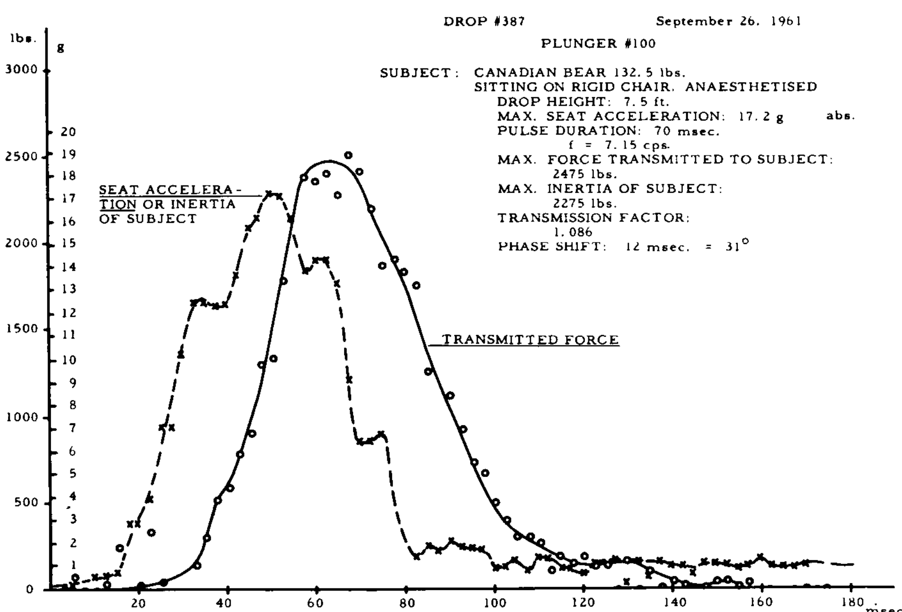

Figure 9. Time history of seat acceleration and transmitted force produced by the vertical deceleration tower.

by the ratio of the maximum transmitted force divided by the mass of the test subject to the maximum of the seat acceleration. The time-distance between the peaks of the two curves related to the time-duration of the pulse was calculated as phase-shift. A rigid weight was first used as the test subject to prove that the chair was stiff enough to obtain a transmission factor of one, and to produce no phase-shift for the range of pulse duration studied.

For the first series of tests only one plunger was used, giving deceleration curves with similar shapes at all drop heights. These deceleration curves can be considered as half-sine pulses, even though high-frequency distortions are superimposed. However, pulses with time durations shorter than 10 ms are not transmitted to the subjects studied, as seen in the force curve. The error will, therefore, not be of significance if frequencies above 100 cps are neglected.

The two curves for "seat deceleration" and "drop height" plotted versus pulse duration in Figure 10 show that the peak of the seat deceleration is proportional to the drop height up to 15 ft., while at 17.5 ft. the maximum deceleration lessens. The pulse duration decreases with increasing drop height, giving shorter but higher pulses. With the plunger used, pulse durations between 46 and 140 ms could be produced. For shorter or longer pulse durations with the same peak deceleration, other plungers must be utilized.

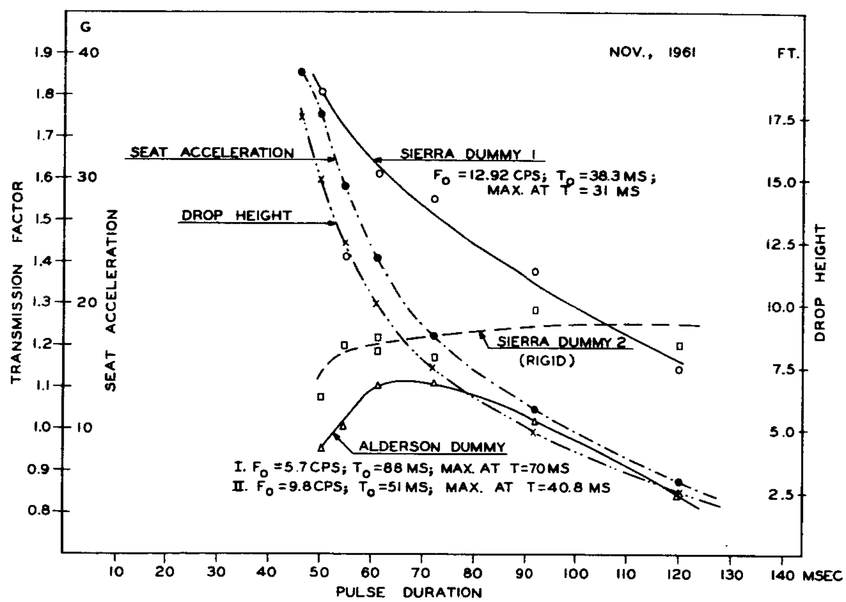

Figure 10. Response to impact of three different dummies.

Preliminary Results

The transmission factors of the three dummies tested are plotted versus pulse duration in Figure 10. Sierra Dummy 1, with its dominating natural frequency at 12.9 cps (natural period t_o = 38.8 ms), should show a maximum transmission factor at pulse duration T = 31 ms (0.8 t_o). This test condition could not be covered by the plunger used, but the trend of the curve points to a maximum in this region. The slight resonance at eight cps (T = 50 ms), may become evident also if more test points are taken.

Sierra Dummy 2 with its rigid structure should show a transmission factor of one, at least down to T = 50 ms. This factor is higher than one above T = 60 ms, and may be explained by the response of the rubber layer used to simulate the skin and outer soft tissues. This layer does not move against the steel structure at low acceleration, and therefore does not change the impedance curve under low-g vibration (see Figure 7). But at higher-g drops the rubber transmits a transient force with high damping with decreasing efficiency at short pulse duration. Generally this curve does not elicit an obvious resonance.

On the contrary, the transmission factor of the Alderson Dummy reveals a resonance between 60 and 70 ms pulse duration. Due to the fundamental resonance of the dummy at 5.7 cps, this resonance had to be expected at T = 70 ms. The second resonance at T = 40.8 ms was not reached by the test conditions.

These measurements indicate that dominating resonances of complex dynamic systems excited by impact can be detected by the described method. The application of this method to living organisms therefore seemed to be justified. Figure 11 illustrates the response to impact of an anesthetized bear and of three human subjects. From Figure 5 the fundamental resonance of the Himalayan bear is shown to be about four cps. A Canadian bear, which was taller in the sitting position than the Himalayan bear and weighed 132 lbs, was used for the impact tests. To keep the bear in upright position during the heavy impacts, it was necessary to restrain his chest to the back of the seat, restricting the downward movements of his skeleton considerably, but not the displacement of the inner organs. These facts may explain that the maximum of the transmission curve for the bear was found for pulse duration below 50 ms. Also the phase-angle seems to cross the 50-degree line at around 40 ms, indicating a resonance at about 10 cps. The enhancement of the transmission factor for 120 ms pulse duration may be related to the movements of the abdominal organs.

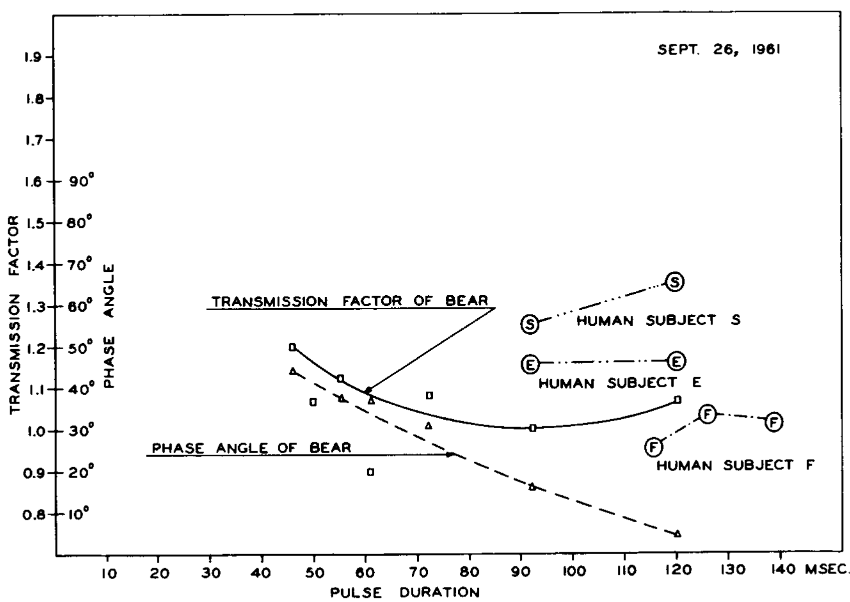

Figure 11. Whole body response to impact of man and Canadian bear, sitting.

The tests conducted with the three human subjects were the first human tests on the drop tower. The subjects did not have any experience with drop tests. They were restrained firmly with standard USAF harness, and instructed to unburden their spine as much as possible by the arm supports of the chair. An impedance measurement of subject F under this condition showed only a very slight peak between five and six cps with high damping. Consequently, it could not be expected that the transmission factor would elicit a distinct resonance.

The few drops conducted were certainly not enough to permit making any conclusions, especially since the shortest pulse duration was longer than 80 ms where the resonance for five cps was expected. An interesting factor is the influence of the body configurations of the subjects. Subjects S and E were relatively short and stout young men, while subject F had about an average body size. This may explain how it is that the transmission factors for S and E are remarkably higher than those of F, since more soft tissue is displaced compared to the bony mass. It may be that this is the reason for the very high transmission factor of S at 120-ms pulse duration. However, these few tests are not enough to establish an impact resonance for humans.

Conclusions

Man, animals, and dummies have different dynamic characteristics which can be determined by measurement of the mechanical impedance under steady-state vibration. While the parameters of the dummies are constant for a wide range of acceleration amplitude and pulse duration, the response of living organisms depends upon the magnitude of the applied force. The theory of impact explains the relationship of whole-body resonances to the response to impacts with varying pulse durations. By measuring the transmitted force to the body during impact, the change of the fundamental frequencies of the test subject can be detected. Transmission-factor and phase-shift are revealing criteria of this test method.

The test series with the bear demonstrates the importance of the restraining system in the response of an organism to transient forces. The few tests with humans show the influence of body configuration on the transmitted force.

Theory and practical tests prove the necessity of determining the mechanical characteristic of the test subject under the actual test condition before any conclusion can be drawn from one test subject and applied to another. Tolerance-limits to impact can be estimated from tests below tolerance if the ultimate strength of the most-stressed organ is known. However, conclusions from animal tests cannot be applied directly to humans if the variations of dynamic characteristics and tissue strength are not taken into consideration.

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A COMPARISON OF THE RESPONSES OF MEN AND DUMMIES TO SHIP SHOCK MOTIONS

Arthur E. Hirsch
Personnel Protection Section
David Taylor Model Basin
Washington, D. C.

Part 1

Tests in which men are exposed to hazardous shock environments are prohibited. With the exception of a few documented cases of accidental exposures to injury-producing accelerations, the bulk of our knowledge of man's response to hazardous shock motions has been derived from dummy or animal work. Dr. Coermann has discussed this problem in some detail in the previous paper. I would like to report some preliminary findings from tests made to determine the validity of the use of anthropomorphic dummies to simulate the response of man to intense accelerations of short duration, such as those met by shipboard personnel during mine or torpedo attack.

The test platforms chosen for these experiments were the destroyer USS FULLAM (DD 474) and the minesweeper MSB-11. These ships were subjected to full-scale mine attacks of varying intensities (see Figures 1 and 2). Deck motions were measured and

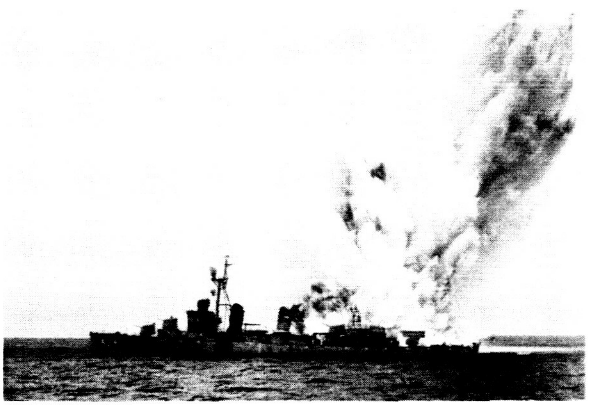


Figure 1. USS FULLAM (DD 474).

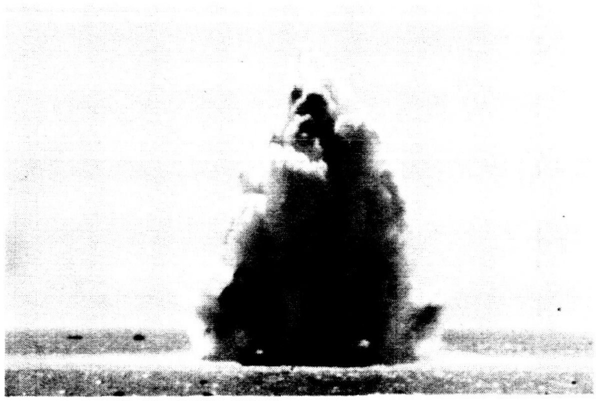


Figure 2. MSB-11.

the the response of the center of gravity of men and dummies to these motions were examined by means of high-speed photography.

A man and a dummy were placed side by side in one of three positions: standing stiff-legged, standing with bent knees, and seated in an uncushioned wooden chair.

The dummies were developed by the Alderson Research Laboratories. A standard F-50 dummy was used for the seated position, a standard C-50 for the bent-knee position, and a modified C-50 dummy in the stiff-legged position. This modification consisted of replacing the rubber pads on the soles of the dummy's feet with a one-inch cut of round steak (see Figure 3).

Figures 4, 5, and 6 are still-shots taken from the motion-picture records.

Figure 3. Feet of stiff-legged dummy with steaks attached.

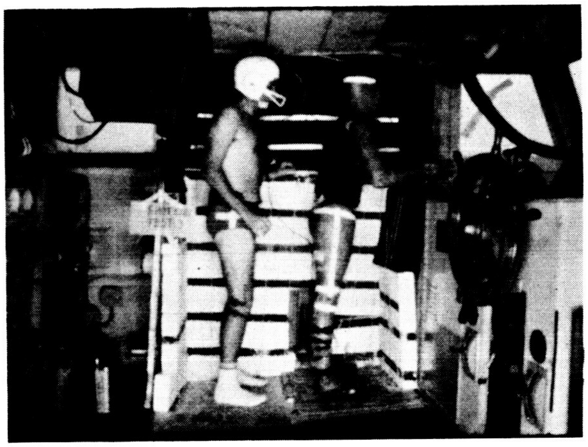


Figure 4a. Stiff-legged man and dummy at zero time.

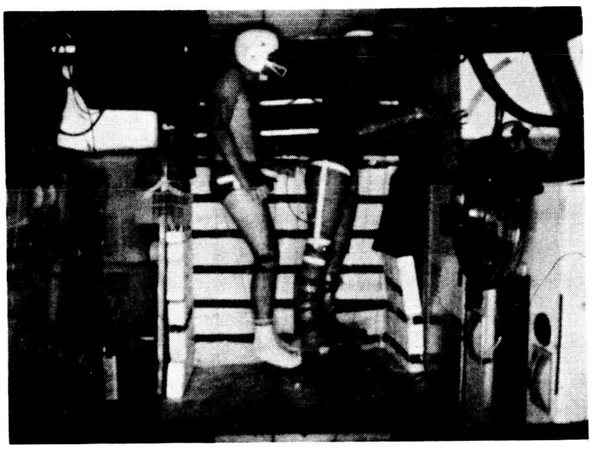


Figure 4b. Stiff-legged man and dummy at maximum displacement.

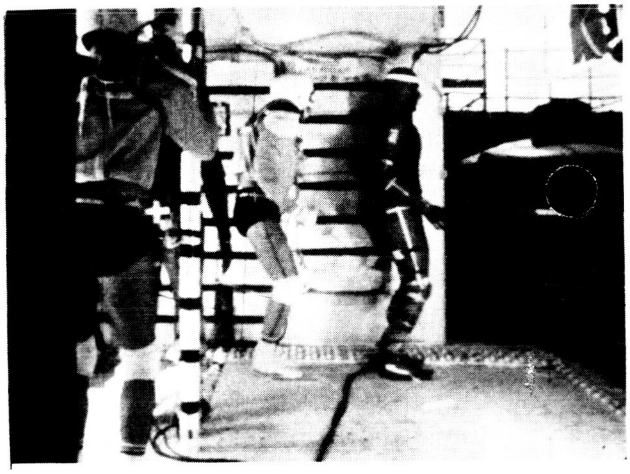


Figure 5a. Man and dummy with bent knees at zero time.

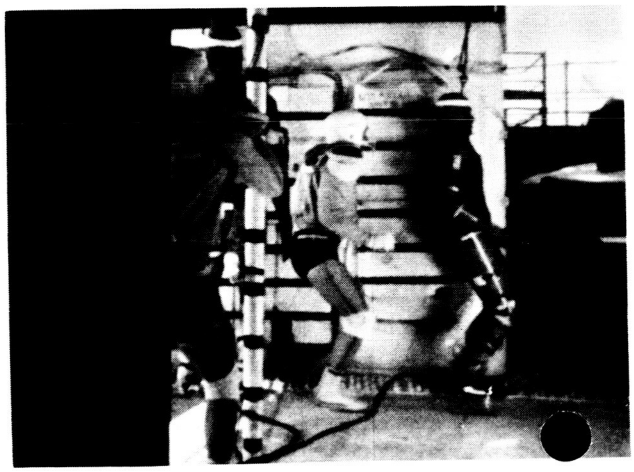


Figure 5b. Man and dummy with bent knees at maximum displacement.

Figure 6a. Seated man and dummy at zero time.

Figure 6b. Seated man and dummy at maximum displacement.

Figures 7, 8, and 9 show the center-of-gravity motions of the subjects in the three positions when exposed to the deck velocities and accelerations also shown in the figures.

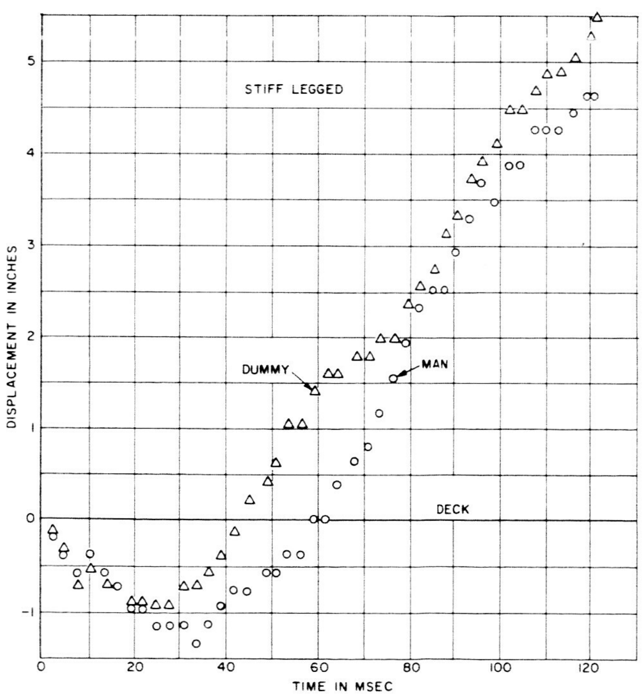


Figure 7. Motion of the C.G. of a standing stiff-legged man and dummy relative to the deck.

As can be seen in Figure 7, the motion of the center-of-gravity of the modified dummy in the stiff-legged stance is very close to that of the man. Figure 8 shows that in the bent-knee stance the dummy does not simulate man's motions. Figure 9 indicates that the motions of the seated man and dummy are not too different. It is possible here that with a better dynamic match of buttock material—for example, if beef steak is substituted for the existing sponge rubber—the results for seated position may be as good as for the stiff-legged stance.

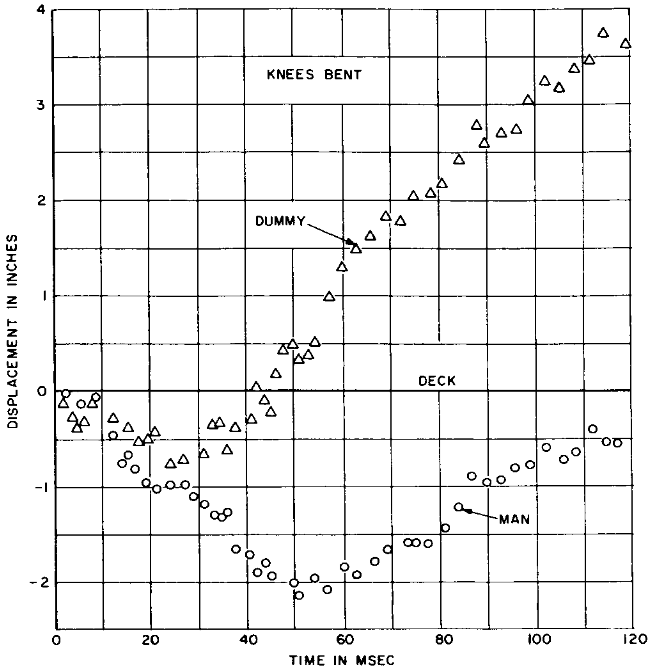


Figure 8. Motion of the C.G. of a standing man and dummy, with knees bent, relative to the deck.

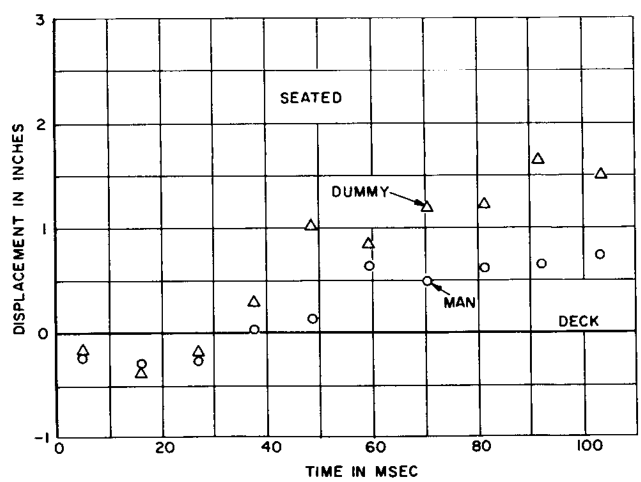


Figure 9. Motion of the C.G. of a seated man and dummy relative to the deck.

Part 2

Comments on the Personnel-Protection Program at the David Taylor Model Basin

The effect of intense short-duration accelerations on man is a subject for consideration by the Navy Bureau of Ships. This concern, as Captain Smith has mentioned, has arisen from a consideration of the vulnerability of ships to underwater explosive attack. Improved hulls and shock-mounting techniques for ship components show promise of significant increase in the tolerance levels of ships to underwater explosion attack. What is the tolerance level for the ship personnel? If it is low, how can it be improved?

An intensive examination of the experimental literature has revealed very little useful data. The problem is that most researchers have been concerned with impacts of long duration and low amplitude. Significant ship-shock motions involve time durations of less than 15 msec and accelerations of greater than 30 g. An additional problem is that the operation of a ship requires that men be relatively unrestrained while seated and have complete mobility while afoot. Thus the mode of entry of the shock to the body is usually up through the feet. The major concern of the researchers in aircraft and space-flight safety is in restrained men in seated or lying positions, where the mode of shock entry is different. Besides the direct shock for unrestrained subjects there is the added hazard of loss of footing and flight which can result in collision and impact injury. This, incidentally, is an area of direct concern to the automobile-safety researchers also.

What is the relative importance of these two injury-producing mechanisms to our work? We have made an analysis of war-damage studies in which ships have undergone underwater explosive attack and have reported details, including injuries. Figure 10 shows a breakdown of these injuries by body area. Since the vast majority

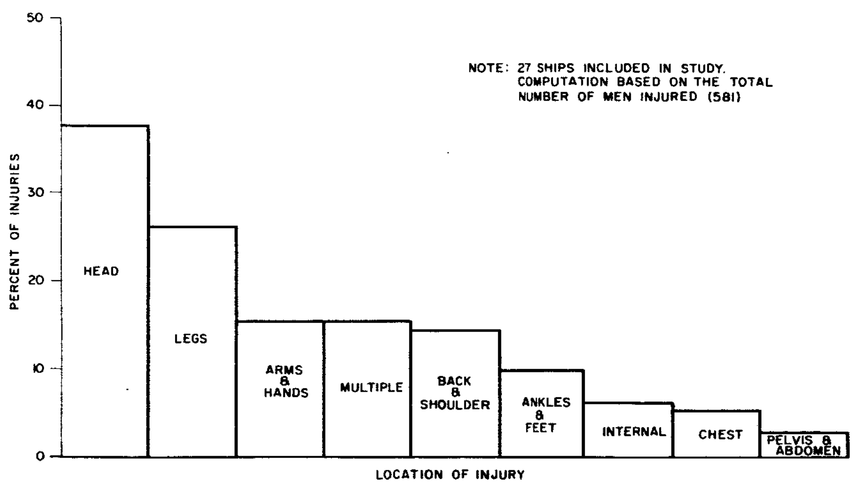


Figure 10. Distribution of personnel injuries, incurred during mine and torpedo attacks, according to body area.

of men on an operating naval ship are standing, it has been assumed that most head and body injuries were a result of collision or missile impact, and foot and lower leg

injuries were caused by direct shock motions transmitted upward through the deck. As can be seen, the major cause of injury is collision; however, there is a significant enough number of injuries from direct shock to warrant attention.

How can we attack these problems? We cannot, as the automobile and aircraft people do, advocate restraints or packaging for any significant number of crewmen.

We can advocate removal of missile hazards, and a softening of the personnel location in some fashion similar to collapsible steering columns, padded dash panels and sun visors suggested by the automobile safety men. A modern fighting ship does not lend itself to this too well, although much of the hardware design could well be modified to serve the same purpose. Sharp corners could be rounded, and exposed structural beams and plating could be coated with a soft material if it were not too thick. The obvious question here is what material and how thick? Would it not be better for men to wear their own impact padding?

This question is being considered. An examination of protective helmets and shock-absorbing materials is in progress, with particular consideration of the ship problem. An experimental program has been underway—and you have seen a part of it in the motion picture I presented this afternoon—to determine the magnitude of the impact a man will sustain under different conditions of ship attack. Coupling this knowledge with tolerance values of skull impact, being developed by our colleagues at Wayne State University, for example and others, a reasonable answer to the question of how much protection, if any, is useful will be found.

For protection against direct shock we must obtain damage criteria from the ejection-seat studies made by the aircraft safety people on injuries during ejections. As usual, it is most difficult to get numbers, although there are ample clinical studies. The most useful data for our purpose are derived from Swearingen's drop tests. We have hopes that the program of investigation of falls being carried on at the Federal Aviation Agency will be helpful. We are planning full-scale laboratory shock tests with a HYGE machine, shock-platform tests, and detailed analyses of war-damage reports to fill in the gaps in our knowledge of man's tolerance to "impact-acceleration stress."

17863

A MINIMAL COMPRESSION FRACTURE OF T-3 AS A RESULT OF IMPACT

Galen A. Holcomb
and
Marilyn Huheey
Stanley Aviation Corporation

The subject is a 22-year old, five-feet 11-inch, 175-pound Mongolian male (Figure 1) in excellent health with virtually no past history except for childhood diseases and asthma to age 12. A complete Air Force, Type 3, Class 1, flight examination was performed on 28 February 1961. All findings were within normal limits and the spinal X-rays showed vertebral bodies and interspaces normal.

The tests for which the subject was eligible consisted of placing the subject in the B-58 escape capsule and simulating ground impact as it would occur during landing on either dry compacted soil or water in high-wind situations. This was accomplished by suspending the B-58 capsule from a monorail and propelling the capsule down the rail at the desired speed by means of an air-pressure device, and releasing the capsule with a quick-release mechanism over the desired type of surface. The capsule's normal landing attitude, which places the man in a pseudo-fetal position lying on his back, was utilized in all biological tests. The orientation about the X axis could be varied so that the subject was traveling down the rails either head forward, feet forward, or side forward.

Each subject underwent complete medical surveillance which consisted of:

1. A physical examination on the morning of the test (blood pressure, pulse, body temperature, heart, lungs, neurological examinations, and urinalysis, as well as examination for scratches, bruises, etc.).

2. Blood pressure, pulse, anxiety, or other subjective reactions, etc. at the test site immediately before and after each test.

3. A repeat of the pre-test examination after removal from the test vehicle.

4. Urinalysis two hours after the test.

5. Repeated pre-test examinations 24 and 72 hours after the test.

Each subject was to refrain from eating or drinking from at least six hours before the test until after the two-hour urinalysis.

This subject performed four drop tests. There were two water drops and two dirt drops. The first was on 2 March 1961 and consisted of a head-forward, 23-mph test from a height of 9'9" onto water. The second was on 7 March 1961, a side-forward water drop (23 mph from 9'9"). The only effects from these drops were two small non-tender bruises on the right thigh and right groin.

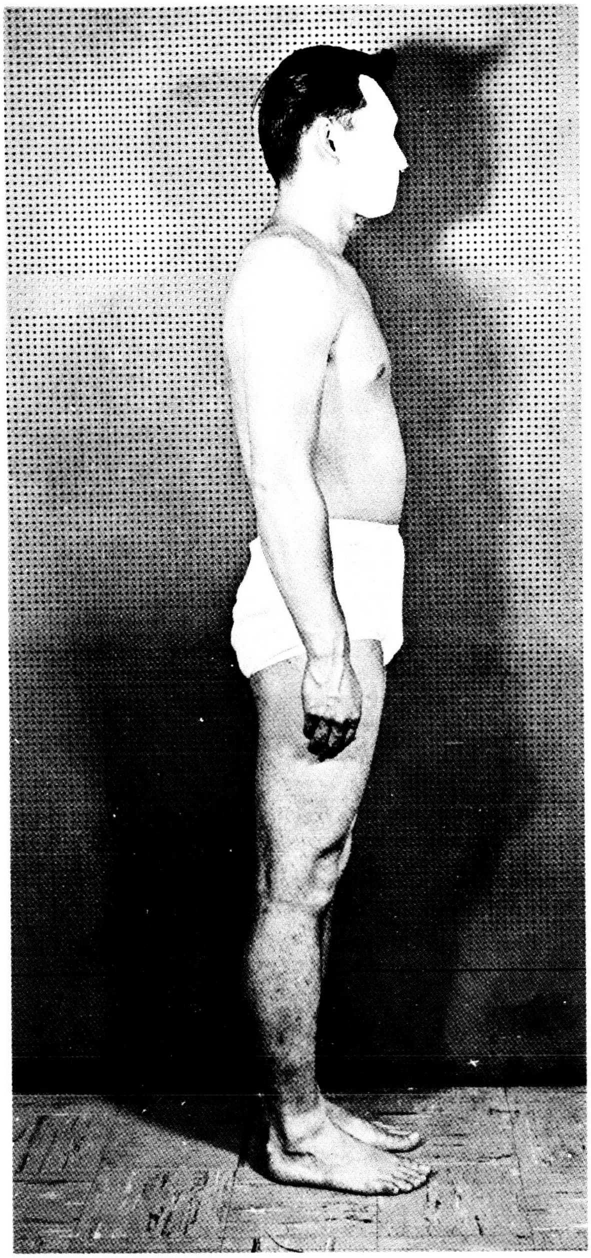


Figure 1. Test subject C.S.—monorail drop tests.

The first dirt drop consisted of a side-forward, nine-mph drop from 9'9" on 6 June 1961. The medical examinations were negative throughout. The last drop was performed by this subject on 29 June 1961, and consisted of a feet-forward, 16-mph drop from 9'9". The subject stated that he had some pain in the area of the left lower scapula—"sharp pain initially—eased rapidly." The doctor reported "no tenderness in area of pain, no evidence of fracture, and no X-rays indicated." The two-hour urinalysis revealed 10-15 WBC/HPF, occasional clumps of WBC's, and a + albumin.

On 30 June 1961, albumin was negative, microscopic examination of the urine revealed 5-6 WBC/HPF, and no clumps. The subject reported "moderate tenderness over the paravertebral muscle to the left of T-3 to T-4—spent restless night with considerable stiffness and discomfort of upper thoracic paravertebral muscles, but feels much better this A.M."

On 5 July 1961, urinalysis was negative and subject reported "pain on left paravertebral area level of T-2 and T-3 on lifting, sneezing, coughing, and certain movements. No tenderness on area." The doctor recommended X-rays of dorsal spine AP and lateral. X-rays were taken on 6 July 1961 and the radiographic report stated, "There is a compression fracture of D-3 with loss in height of the centrum amounting to about 4 mm. Comparison with films made 28 February 1961 show that this change was not present on the early examination."

In a drop test of this type, it is easily seen that the acceleration experienced upon impact is not ever a simple force in the spinal, transverse, or lateral directions. Changes in attitude and vertical and/or horizontal speeds change not only the magnitude of the acceleration but the direction as well. Therefore, no attempt will be made to point out that there were other tests in which higher transverse or spinal accelerations occurred, or that higher resultant accelerations were experienced by other subjects during the same test program without ill effects. The direction of this force has been determined to be approximately 46 degrees from the X and Z axes, and approximately 10 degrees laterally. (Figure 2).

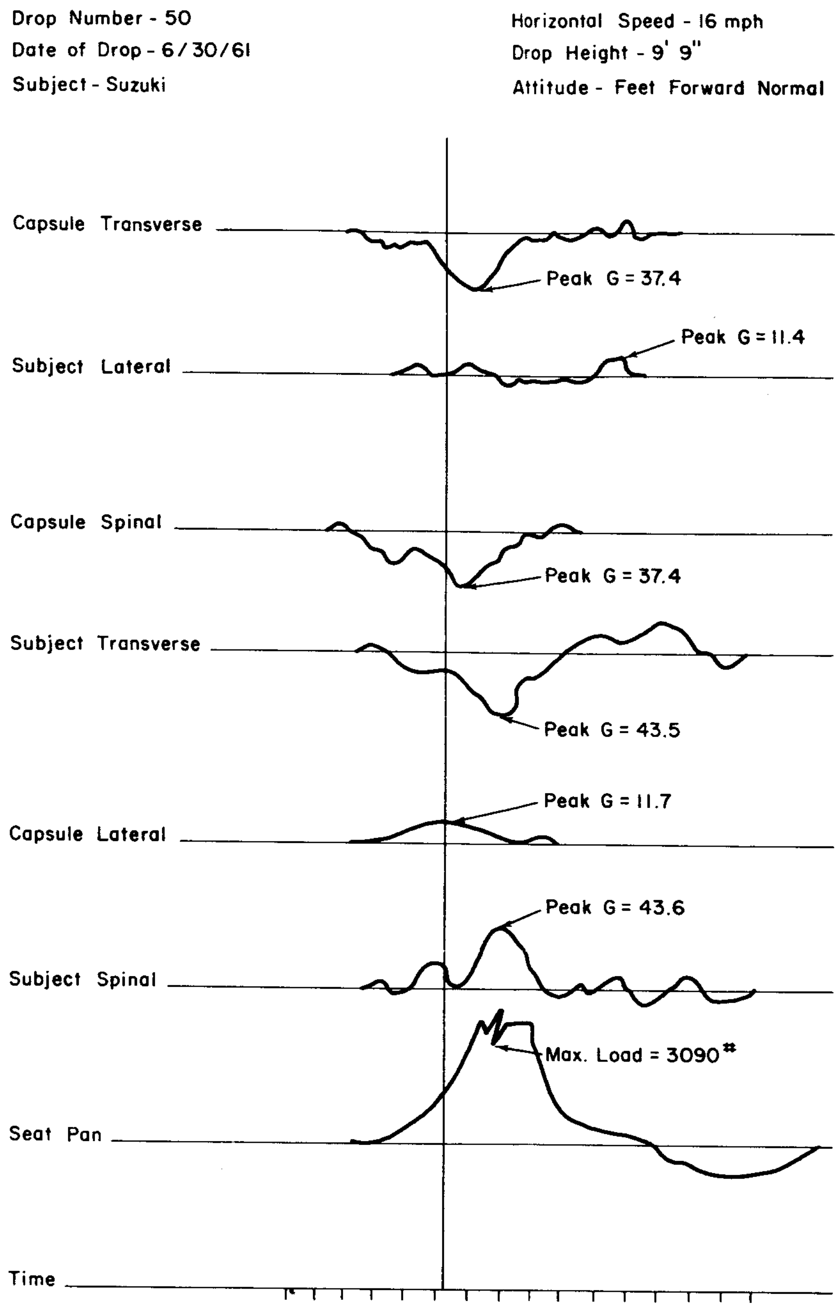


Figure 2. Copy of actual acceleration trace. From the capsule X, Y, and Z traces, resultant magnitude and direction are computed.

The possible effects of a force with these particular characteristics is illustrated in Figure 3. As the capsule impacted, the vertebrae tended to move toward the seat back. The cervical and first two thoracic vertebrae are nearly parallel with the direction of the force. T-4 was probably impinged upon the edge of the seat-back cushion. Therefore, with the upper vertebrae moving toward the back and buttocks, T-3 could have been mildly compressed such that a permanent loss in height of the centrum of about 4 mm occurred.

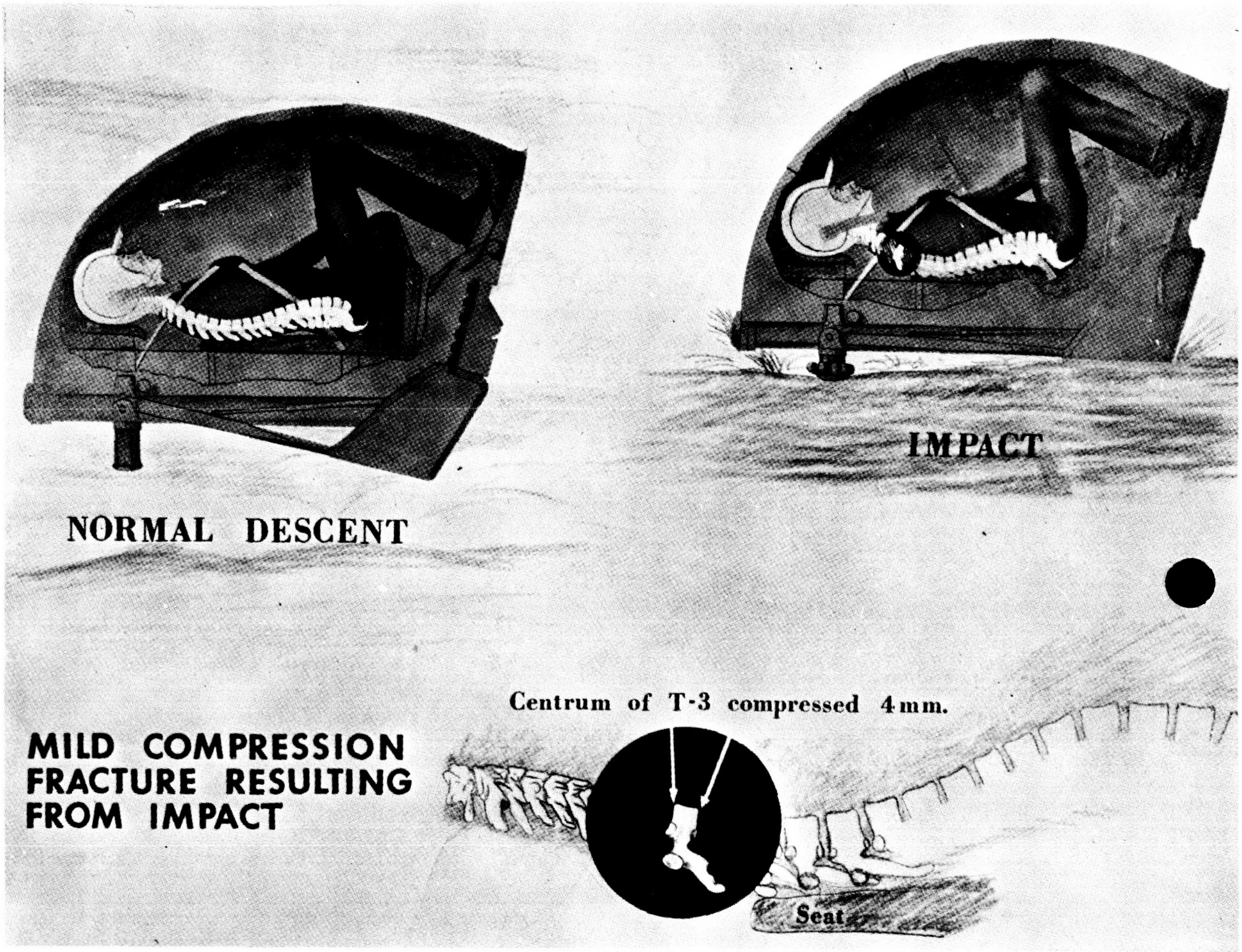


Figure 3. Illustration which represents position of occupant during descent and upon impact. Lower portion represents the affected vertebrae in close up.

Another possible event contributing to this compression was that the neck muscles (the sternocleidomastoid, in particular) were probably tensed and putting another force on the upper vertebrae. The rib cage could have flexed while the vertebrae moved toward the seat back and caused a combination of stretching and compressing of the interspaces such that the centrum of T-3 was compressed. Other effects were probably also present.

Thus, even though the occupant of the B-58 capsule (as well as all presently planned U. S. space vehicles) is positioned to receive backward-facing transverse accelerations upon ground impact, the possibility of a spinal injury should be admitted, due to the complexity of the ground impact force.

12560

THE DYNAMICS OF HUMAN RESTRAINT SYSTEMS

Peter R. Payne
Frost Engineering Development Corporation

Introduction

Human dynamics is in its infancy, and, like all young sciences, must proceed in a series of steps which alternate between theoretical and experimental investigations.

In the sub-division of body-restraint dynamics a great deal of experimental data has now been amassed, and further progress seems to depend upon a thorough investigation of the basic principles of restraint, and the use of dynamic theory to correlate the existing experimental information. This report is primarily concerned with proposing such a program, and with discussing in depth the approaches that should be used.

Since a satisfactory dynamic model of the human body is an essential prerequisite to a meaningful analysis of restraint dynamics, however, a fairly detailed description of our latest "human models" is also included.

A Revised Dynamic Model Concept

The dynamic models originally proposed by the writer(12,8) could be used to predict the limits of human tolerance if it was assumed that there was a critical peak mass acceleration $\ddot{y}_{P_{MAX}}$ which marked the boundary of tolerability.

This assumption was taken directly from the work of Galen Holcomb, who has maintained for some years that subjective response should be correlated with the output rather than the input acceleration. Since in practice the human body is too complex a dynamic system for an accelerometer to give consistent correlation for all types of input, we are forced to calculate the output of an "ideally mounted" accelerometer mathematically.

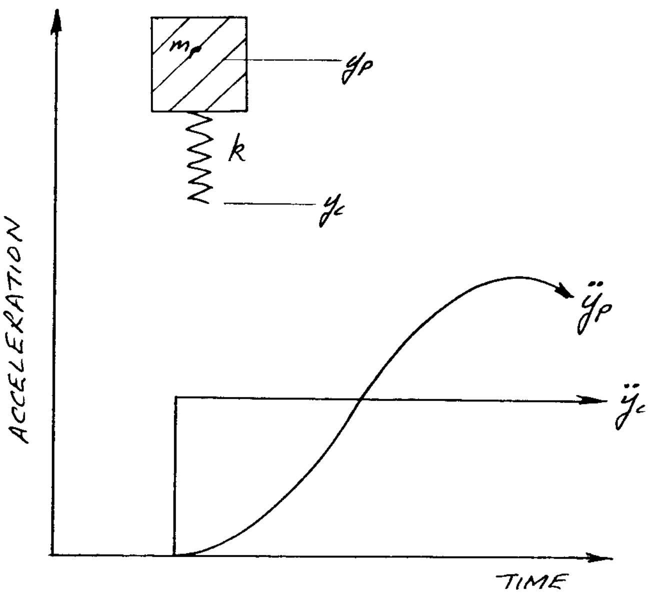


Figure 1

In effect, the assumptions used imply the existence of a critical peak load or force in the spring, since the damping was not considered, and the model could therefore be rationalized in terms of an easily understood physiological limitation. For example,

we know that the most common physiological limit to acceleration in the spine is an anterior compression fracture of one or more of the vertebral bodies in the lower thoracic-upper lumbar region of the spinal column. The mechanism of this failure and the physical variations which can effect it are described by Bosee and Payne in Reference 21, where it is shown that both seat design and the posture of the subject can cause wide variations in the tolerable load in the spine. Thus we must tie our theoretical failing load to a definite posture, and in our present state of ignorance it is most logical to assume that we are dealing only with properly supported subjects.

With the continuation of work in the field of human dynamics it has become necessary to include damping terms in our mathematical representation of the human body. A very practical reason for this is that arbitrary acceleration-time histories can be economically studied only on analog computers, and an analog circuit with zero damping is highly undesirable. A second and more important reason is that the motion of the human body is damped, and although we are currently unable to assign a numerical value to its damping coefficient, at least with any pretense of accuracy, it is better to make a "best guess" than to ignore the subject entirely. Our current "best guess" is a critical damping ratio of 13 per cent when the load is close to the tolerable limit, increasing to as much as 60 per cent for small amplitude deflections of the type studied by Dr. von Gierke, Coermann, et al at ASD. [7]

When damping is included we have to decide whether our limiting parameter is the spring strain or the total force induced in the system, since the "spring force" and the "total force" now differ by an amount equal to the load in the damper. If we were to retain Holcomb's original postulate, we should of course settle for the total load as our "Physiological Index." In the spinal case at least, however, there are some reasons for believing that the "spring force" comes mostly from the lightly damped bone structure, and that the damping elements are in parallel with it. Thus in this case we should settle for the peak spring strain as our "Physiological Index."

This "critical strain" therefore appears to be more in accordance with physical reality. If we again consider the case of spinal acceleration, for example, the early work of Ruff[17] et al, in Germany, tells us the compressive strength and stiffness of the average human spine, and the mass associated with it. This information enables us to calculate the input acceleration which will just cause a compressive fracture of the average spine, and to plot a tolerance limit as shown in Figure 2. Such a calculation assumes that all the body's mass is supported by the spine, and a curve of the type shown in Figure 2 formed the basis for many early ejection-seat specifications.

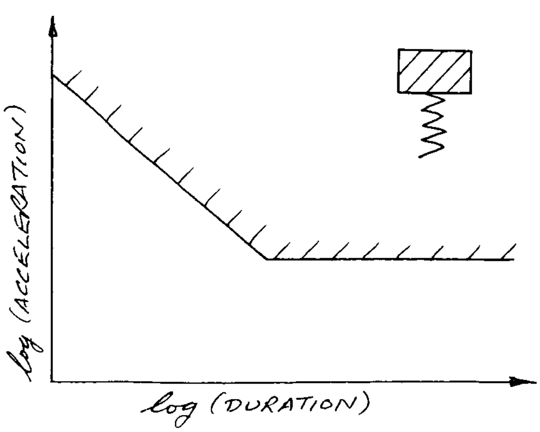


Figure 2. Calculated Tolerance of Spine in Compression

Whilst considerations of mechanism are often instructive, it should always be remembered that our primary purpose at the present time is to produce an equation, or a set of equations, which adequately explain all the available experimental data in the light of some simple dynamic hypothesis. Any reference to what is actually happening in the test subject's body is irrelevant to the consideration of gross phenomena, and the very complexity of the body's reaction may in fact mislead us in our

analysis. What mathematics provides in this case is a logical measuring scale, against which physiological phenomena can be correlated. The sole function of the mathematician-dynamicist is to provide this measuring scale, convince research workers in the field that it is valid, and then withdraw to the wings.

Naturally the mathematician will have to be called upon from time to time, to keep the measuring scales up to date as new information is generated. He also has to provide instruction on occasion, since medical facilities do not normally include mathematical dynamics in their curriculum.

The rest of this section is devoted to discussion of some elementary properties of a dynamic system, and for simplicity we consider only the case of zero damping. Finite damping is introduced subsequently in the section on Dynamic Models of the Human Body, in considering developments of specific dynamic models.

When a simple undamped dynamic system of the type shown in Figures 1 and 2 is subjected to a rectangular acceleration pulse of the type defined in Figure 3, then the peak load in the spring (and therefore the maximum acceleration of the mass) is a function of the duration Δt over which the acceleration $\ddot{Y}_c$ is applied. In particular there is a critical duration Δt_1, which marks the dividing line between two quite different regimes of dynamic behavior.

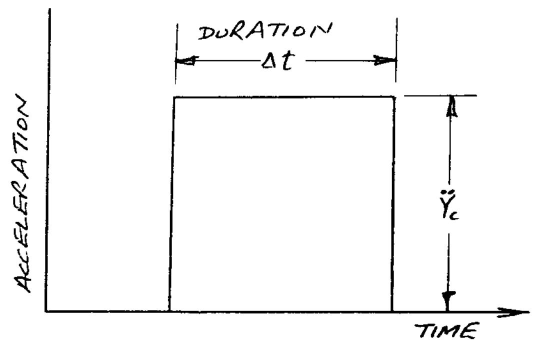


Figure 3. Rectangular Acceleration Pulse

"Impact" Accelerations

If the duration of an acceleration pulse is less than this critical duration Δt_1, then we are dealing with an impact problem, in that the pulse is over before the mass has had time to react to its influence. We may write this mathematically by saying that the actual value of Δt_1 depends on the natural frequency of the dynamic system, the actual relationship being approximately

$$\Delta t_1 = \frac{2}{\omega} \qquad (1)$$

where $\omega = \sqrt{k/m}$ radians/sec.

natural frequency $= \frac{\omega}{2\pi}$ cycles/sec.

For impact accelerations ($\Delta t < \Delta t_1$) neither the duration nor the magnitude of the acceleration are significant, but only their product ($\Delta t \ddot{Y}_c$) which is the velocity change Δv imposed upon the system.

For this reason it is easy to calculate the peak spring strain in the impact range by equating the kinetic energy of the system before impact to the potential energy stored in the spring after impact, since all the kinetic energy stored in the spring by the impact must momentarily appear as potential energy before rebound converts it back into kinetic energy.

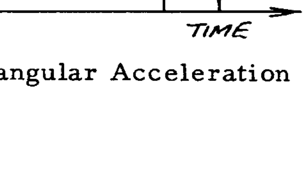


Figure 4

Thus $\frac{1}{2} m \Delta v^2 = \frac{1}{2} k \delta_{MAX}^2$

$$\therefore \quad \delta_{MAX} = \Delta v / \omega \tag{2}$$

where $\omega^2 = k/m$

and the maximum spring force $k\delta_{MAX}$ =

$$F_{MAX} = k \Delta v / \omega \tag{3}$$

In general it is obviously more convenient to use the spring deflection as the basic measure of physiological effects, rather than the force, since it does not require a knowledge of the effective spring rate k, but only of the natural frequency ω. It should be noted that the assumption of a linear model without damping results in equivalent spring strains δ_{MAX} which are obviously too large to occur in the real system, due to the fact that the real system is both non-linear and has damping. This does not affect the validity of their use, however, since we are only interested in the values of the equivalent force $k\,\delta_{MAX}$.

The concept of defining the mass acceleration as the critical parameter also conforms easily with this new approach, since

$$\ddot{y}_{P_{MAX}} = \omega^2 \delta_{MAX} \tag{4}$$

"Long Duration" Accelerations

If the duration of the acceleration pulse is greater than Δt_1, then the response of the spring mass system is quite different from its impact response. A "long duration" is defined as being greater than the natural period of the spring mass system ($\Delta t > 2/\omega$). If the acceleration input to a single degree of freedom system is of the form shown in Figure 5, then the mass will "overshoot" the input acceleration by 100 per cent; a phenomenon which is familiar to us from the sudden application of loads in structural stress calculations.

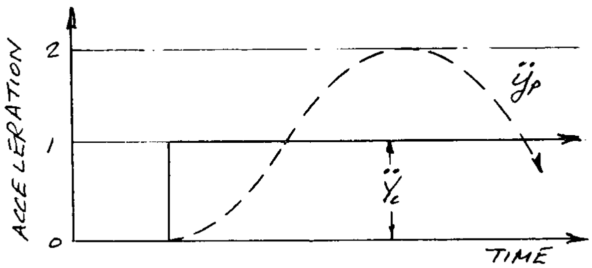


Figure 5. Dynamic System's Response to "Long Period" Acceleration with Zero Rise Time

From equation (4)

$$\delta_{MAX} = \frac{2\ddot{Y}_c}{\omega^2} \tag{5}$$

Thus for "long duration" accelerations, the peak strain δ_{MAX} is independent of the duration of the acceleration, so that we shall not expect the tolerable limit to vary with duration, until other effects (such as blood flow) intervene.

The pulse shape of a long duration input has a significant physiological effect, however, the simplest example being given by the linear shape shown in Figure 6.

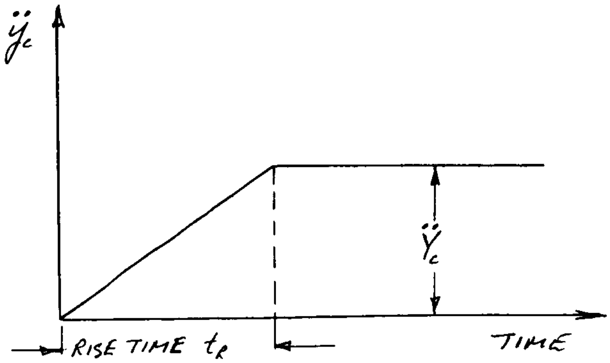


Figure 6. Definition of Rise Time

If the rise time is short, in relation to the natural period of the dynamic system, 100 per cent overshoot will occur, as indicated for zero rise time in Figure 5.

If the rise time is long, then the overshoot is small, so that the same plateau level $\ddot{Y}_c$ of the input acceleration has less physiological effect, and causes less strain or deflection of the human "spring."

The theoretical analysis of a linear rise time is presented in Appendix A of this proposal. In general the maximum spring strain can be plotted as a function of $\ddot{Y}_c$, the plateau acceleration, and the non-dimensional rise time parameter ωt_R. For zero initial conditions the results of this analysis can be plotted as shown in Figure 8. For practical purposes we may assume zero rise time theory to apply if

$$t_R < \frac{1}{\omega}$$

and at the other extreme, we can neglect dynamic overshoot if

$$t_R > \frac{10}{\omega}$$

These results are of considerable importance since they indicate that, if a certain acceleration input is critical when the rise time is long, only half this value can be tolerated if the rise time is less than $1/\omega$.

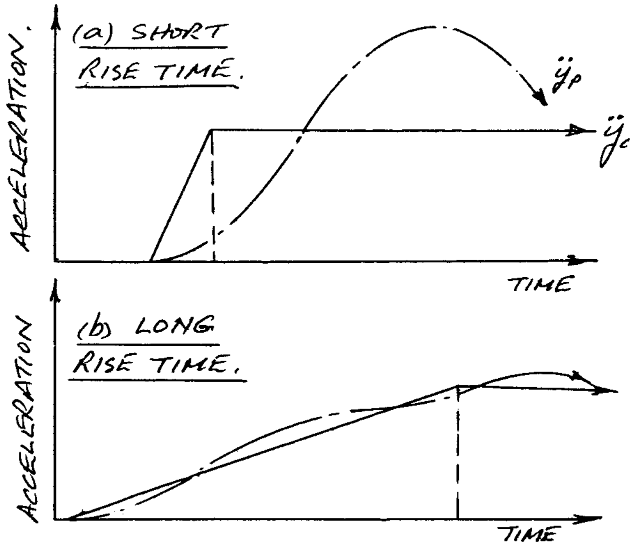


Figure 7. Effect of Rise Time upon Acceleration of Mass

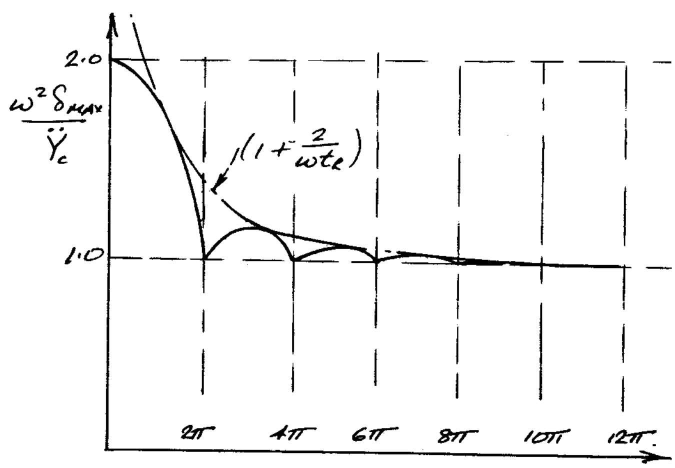


Figure 8. Theoretical Effect of Rise Time

A second result can be seen by considering spinal accelerations in the duration range of .01 to 0.1 seconds, where the appropriate natural frequency of the human body is 278 rads/sec.

A rise time of

$$t_R = \frac{5}{\omega} = .018 \text{ Seconds}$$

would result in only 20 per cent dynamic overshoot, from Figure A.3. (Appendix A)

If we now add a cushion (such as a net seat, for example) in series with this, we could reduce the effective frequency to say 50 rads/sec. The same acceleration input, with a rise time of 0.18 seconds will now give $\omega t_R = 50 \times .018 = 0.9$

resulting in an essentially 100 per cent dynamic overshoot. Thus the addition of the cushion resiliency can result in a system which is physiologically much more damaging for the same acceleration input.

It should be noted that we have only considered a "linear rise time" analytically. In practice we obviously have an infinite variety of acceleration pulse shapes, where the dynamic response will vary in detail from the simple theory given. The chief merit of the mathematical approach to human tolerance lies in the fact that a satisfactory dynamic model must automatically allow for the most complicated pulse shape, and hence eliminates the need to consider as separate variables such parameters as "rate of onset." All of these effects are included when a natural frequency and a critical spring strain have been defined.

Critical Spring Strain in a Two Degree of Freedom Model

The work of Ruff[17] in determining the average strength of the human spine was referred to in the beginning of this discussion. He determined the mass associated with the spine's resiliency by measuring its static deflection, utilizing x-ray techniques, and then calculated the acceleration which would be required to cause a failure, as a function of duration.

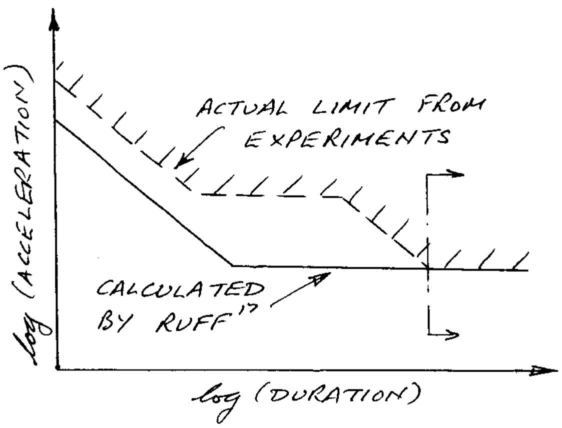


Figure 9. Tolerable Rectangular Acceleration Pulse in the Spinal Direction

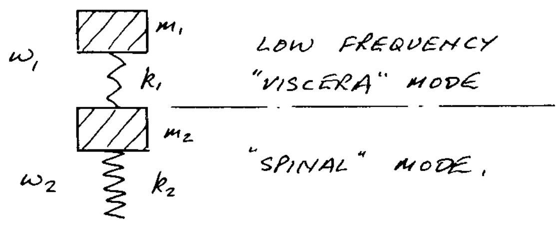


Figure 10. Two Degrees of Freedom Model for Spinal Accelerations

For short duration accelerations, however, the soft, low frequency parts of the body (such as the viscera) do not deflect far enough to load up the spine, and thus the effective mass is lower than the static value assumed by Ruff, resulting in higher tolerable accelerations than those he calculated. If the acceleration duration is long enough for all the body components to deflect, then Ruff's calculations are in excellent agreement with physiological limits established by rocket sled and other methods of testing.

These results can be represented by a two degree of freedom spring mass system of the type shown in Figure 10. Here the upper, low frequency system can load up the "spinal mode" spring if the acceleration endures long enough,

i.e., if the duration

$$\Delta t > \frac{2}{\omega_1}$$

The advantages of defining tolerability in terms of a critical spring strain are now evident, since we are using a single criterion which has a recognizable relationship to the true physical picture, instead of having to define as many critical accelerations $\ddot{y}_{P_{MAX}}$ as there are masses in the system.

The equations governing a two degree of freedom system are solved in Appendix B for impact and long-period accelerations.

Dynamic Models of the Human Body

Although the similarity between the subject of human tolerance to short period accelerations and the classical concepts of dynamics had already been pointed out by Ruff[17], Kornhauser[9,10], Brick[11] and others, it is believed that the possibility of reducing all of the available experimental data by means of a dynamic model, and of using this model to assess the tolerability of any arbitrary acceleration-time history, was first pointed out by the writer[12,8]. This early work has now resulted in a NASA-funded research program at Stanley Aviation Corp., in which an attempt is being made to construct dynamic models which will explain, not only the limits of tolerability, but also the variation of mechanical impedance of the human body with amplitude.

The NASA research program can be regarded as having three phases.

Phase 1 - which is concerned with the construction of linear dynamic models which explain the physiological tolerance data readily available.

Phase 2 - in which an attempt is being made to collect all appropriate experimental data in existence.

Phase 3 - will involve the correction and extension (if necessary) of the original linear models where the new data indicates the need, and an attempt to enlarge the usefulness of these models to problems in which less than critical accelerations are involved, by the introduction of appropriate non-linearities.

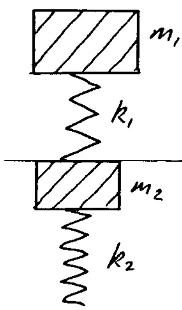
Provisional Definition of Physiologically Tolerable Accelerations

A survey of the experimental data readily available (i.e., without searching the files of the various military agencies conducting physiological experiments) has resulted in the construction of the tolerance limits depicted in Figures 11 and 12, for spinal and transverse accelerations, when the acceleration input is in the form of a rectangular or "square wave" pulse with zero initial conditions, and when the head restraint is "soft." In general, the boundaries are slightly conservative, and mark the threshold at which "healthy young men" start to experience injury. The very limited experimental data available makes it difficult to detect any variation due to acceleration vector sign (such as "eyeballs in" as against "eyeballs out"); thus, these provisional graphs are intended to apply in either direction, provided that head acceleration tolerance is considered as a separate problem, as discussed in the section on Safety Nets.

The dynamic models which give the results of Figures 11 and 12 are as follows:

Spinal Model

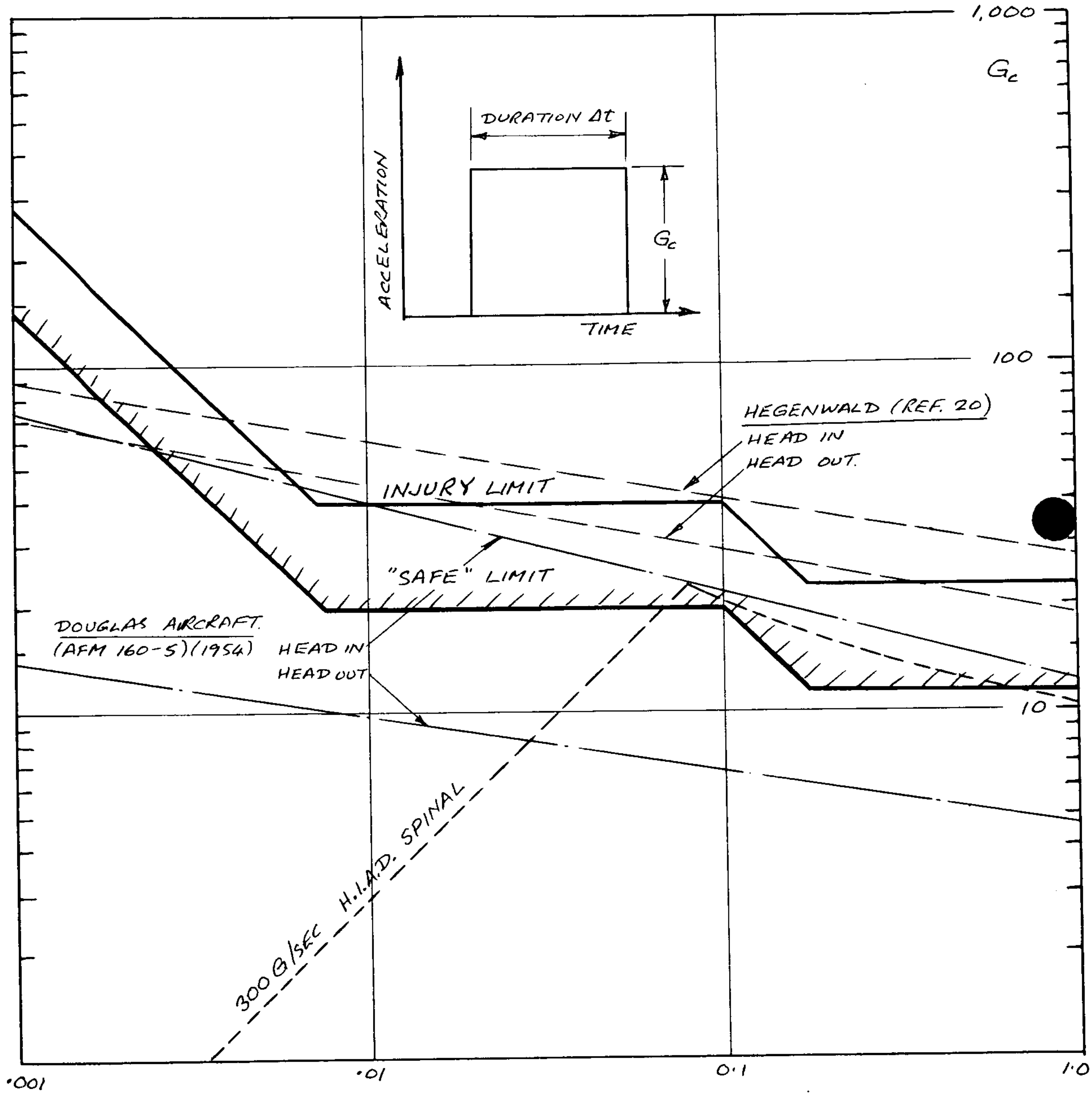
"Viscera Mode"



$\omega_1 = 11.8$ rads/sec
$= 1.88$ cps
$m_1/m_2 = 1.24$

"Spinal Mode"

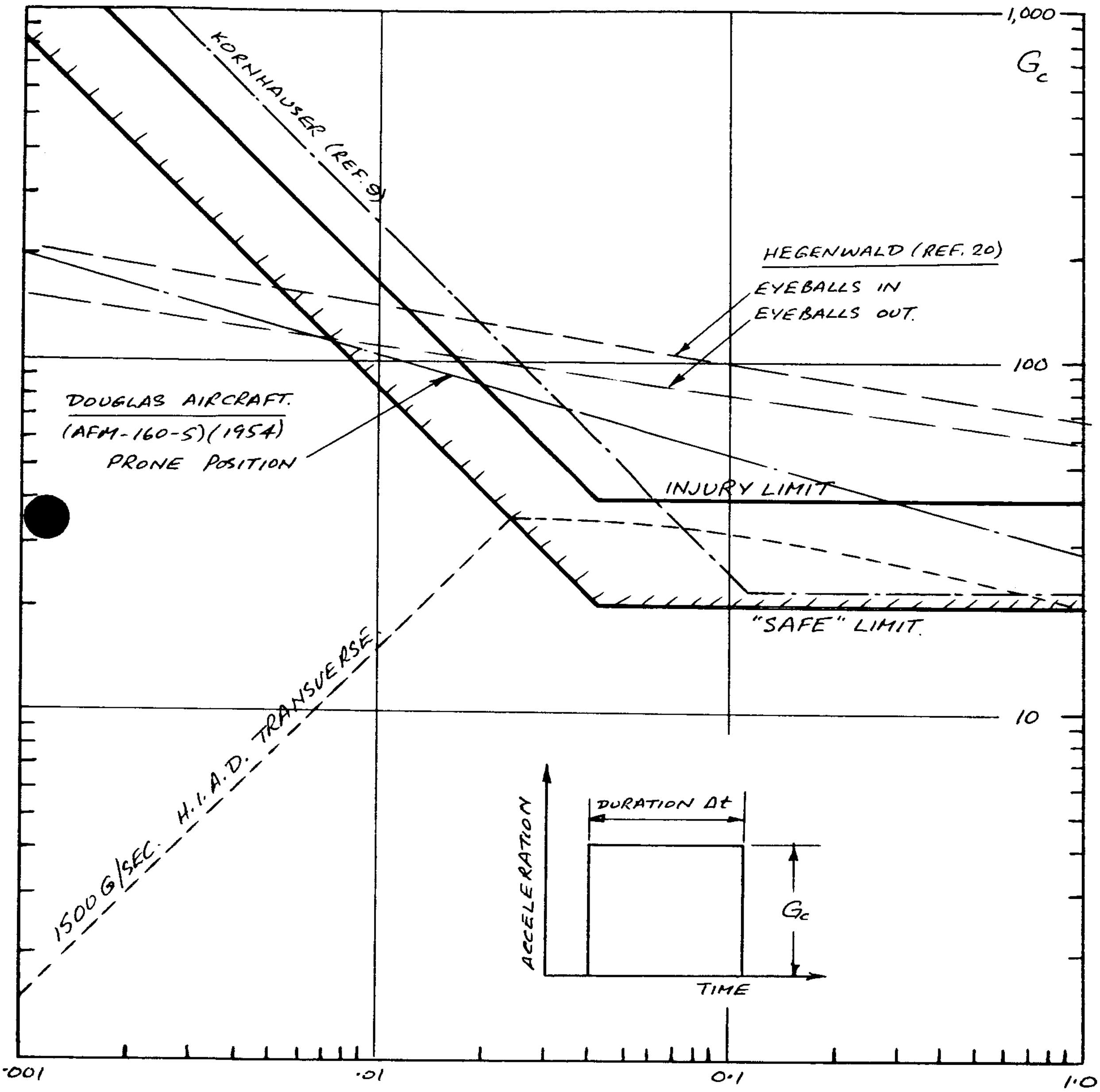
$\omega_2 = 278$ rads/sec
$= 44.3$ cps
$\delta_{CRIT} = .0333$ ft.
$= 0.4$ ins.
$\omega_2^2 \delta_{CRIT} = 80g$



DURATION OF ACCELERATION IN SECONDS - Δt

Note that the HIAD curve is based upon a trapezoidal definition, and therefore is more severe than appears from this plot.

Figure 11. Human Tolerance to Spinal Acceleration with "Soft" Head Restraint



DURATION OF ACCELERATION PULSE IN SECONDS - Δt

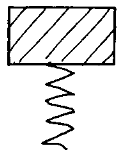
Note that the HIAD curve is based upon a trapezoidal definition, and is therefore more severe than appears from this plot.

Figure 12. Human Tolerance to Transverse Acceleration with "Soft" Head Restraint

The significant parameters derived from these results, for rectangular acceleration inputs, are as follows:

	Viscera Mode	Spinal Mode
"Corner duration" $\Delta t_1 = 2/\omega$	0.17	.0072 (secs)
Critical velocity change for $\Delta t < \Delta t_1$	126.0	9.27 (ft/sec)
Critical short duration input for $\Delta t > \Delta t_1$	23g	40g (ft/sec^2)

Transverse Model



ω = 48 rads/sec.
= 7.65 cps
δ_{CRIT} = 1.118 ft.
= 13.4 ins.
$\omega^2 \delta_{CRIT}$ = 80g

The significant parameters derived from this model for rectangular acceleration inputs are as follows:

"Corner duration" $\Delta t_1 = 2/\omega =$	.0417 (secs)
Critical velocity change for $\Delta t < \Delta t_1$	53.6 (ft/sec.)
Critical long duration input for $\Delta t > \Delta t_1$	40g (ft/sec^2)

Lateral Model

In the absence of any experimental data, and since $\omega^2 \delta_{CRIT}$ = 80g for both spinal and transverse directions, we can tentatively assume that the same value will be applicable to the transverse case when the subject is adequately supported.

We can also deduce an appropriate frequency from the transverse data, by applying a correction based upon the geometry of the average torso, but since this would yield a frequency less than the transverse value (thus yielding a higher critical velocity change for $\Delta t < \Delta t_1$) it seems best to adopt the conservative approach of assuming all the lateral dynamic system coefficients to be the same as for transverse.

Head Model

While the foregoing models would be adequate for a fully restrained man, if the subjects of the experiments on which the models are based were fully restrained, it now appears that head tolerances should be treated separately. This is because the head was free to move forward in the transverse ("eyeballs out") acceleration tests, while the spinal experiments were limited by compressive failure of the spine.

The subject of head tolerance is discussed in Section 5 in some detail, but the general conclusion is that, although it can withstand a 40g "long duration" input acceleration, its natural frequency is higher than for the rest of the body, (20 cycles/sec as against 8 cycles/sec for body transverse) so that it cannot tolerate as high a velocity change under impact acceleration conditions.

Thus a back-landing impact on a rigid couch could be less tolerable than an eyeballs out impact where the head is free to swing forward.

These results are complicated by the fact that in most "eyeballs in" experiments the subject wears a protective helmet and his couch or head support has some resiliency, so that the effective natural frequency of his head plus restraint is lower, and the critical velocity change correspondingly higher. Thus the results of these experiments must be corrected by the use of restraint dynamic theory if they are to be of general utility.

Provisional Definition of "Physically Safe" Accelerations

After a study of the limits of voluntary tolerance, it was decided that a 100 per cent reserve allowance (safety factor of 2.0) would be adequate to cover both the scatter between different (healthy) individuals and the "factor of ignorance" occasioned by the scanty experimental data available. Thus the "specification limits" have been obtained by limiting the theoretical strain δ_{MAX} to half the value at which damage occurs. It is interesting to note that this procedure results in a "specification limit" for transverse which falls below HIAD curves over an important part of the impact duration range. Thus although the transverse HIAD curve is obviously far too low for impact durations of less than about 0.02 seconds, it appears to be too high in the range of 0.03-0.3 seconds; particularly if the head restraint is "hard."

The Effect of Damping in a Linear Model

The dynamic models described in the earlier parts of this section did not include the effects of damping, and a more complete description is provided by the model shown in Figure 13.

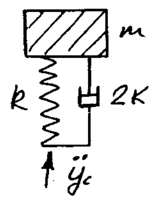


Figure 13. Linear Model with Damping

If $\omega^2 = k/m$

$c = K/m$

then the equation of motion for this model is

$$\ddot{\delta} + 2c\dot{\delta} + \omega^2\delta = \ddot{y}_c \qquad (6)$$

for a constant acceleration input

$$\ddot{y}_c = \ddot{Y}_c$$

With zero rise time, the subcritical solution to equation 6 is

$$\delta = \frac{\ddot{Y}_c}{\omega^2} - \frac{e^{-\bar{c}\omega t}}{\omega^2}\left\{[\ddot{Y}_c - \omega^2\delta_o]\cos\lambda t + \frac{\omega}{\lambda}[\bar{c}(\ddot{Y}_c - \omega^2\delta_o) - \omega\dot{\delta}_o]\sin\lambda t\right\}$$

$\bar{c} = c/\omega = K/\sqrt{mk}$

$\delta_o = \delta$ at time t = 0

$\dot{\delta}_o = d\delta/dt$ at time t = 0 (7)

$\lambda^2 = \omega^2 - c^2 = \omega^2(1-\bar{c}^2) > 0$

The force in the system is

$$F = k\delta + 2K\dot{\delta} = m\ddot{y}_p \tag{8}$$

It can be shown that the peak value of this force is

$$\frac{F_{MAX}}{m} = \ddot{Y}_c - (\ddot{Y}_c - \ddot{y}_s)e^{-\bar{\zeta}\theta_{MAX}} \tag{9}$$

where $\eta = \lambda/\omega = \sqrt{1-\bar{\zeta}^2}$

$$\theta_{MAX} = \frac{n\pi - 2\sin^{-1}\eta}{\eta}$$

and $\ddot{y}_s$ = initial static acceleration at time t = 0.

Also, it can be shown for an impact acceleration in the region $(\Delta t < \Delta t_1)$ that a velocity change Δv will induce a peak force of

$$\frac{F_{MAX}}{m} = \omega\,\Delta v\, e^{-\bar{\zeta}\phi_m} \tag{10}$$

where $\phi_m = \dfrac{\pi - \sin^{-1}\eta - \sin^{-1}2\bar{\zeta}\eta}{\eta}$

The strain δ in the system is no longer proportional to the total force, as in the case with an undamped system, however, since the damper now supplies some of the force.

For the case $\delta_0 = \dot{\delta}_0 = 0$ equation 7 can be written

$$\frac{\omega^2\delta}{\ddot{Y}_c} = 1 - e^{-\bar{\zeta}\omega t}\left\{\cos\lambda t + \frac{\omega\bar{\zeta}}{\lambda}\sin\lambda t\right\} \tag{11}$$

and

$$\frac{\dot{\delta}}{\ddot{Y}_c} = \frac{e^{-\bar{\zeta}\omega t}}{\lambda}\sin\lambda t \tag{12}$$

The peak spring deflection δ_{MAX} will be reached when $\dot{\delta} = 0$,

i.e., when $\lambda t = \pi$, $t_{MAX} = \pi/\lambda$ (13)

Substituting these conditions in equation 11

$$\frac{\omega^2\delta_{MAX}}{\ddot{Y}_c} = 1 + e^{-\frac{\bar{\zeta}\omega\pi}{\lambda}} = 1 + e^{\pi\bar{\zeta}/\eta} \tag{14}$$

Thus the inclusion of damping has led to differing criteria for the physiological effect of a zero rise time acceleration, depending on whether we use spring strain or force. These differences are illustrated in Figure 14.

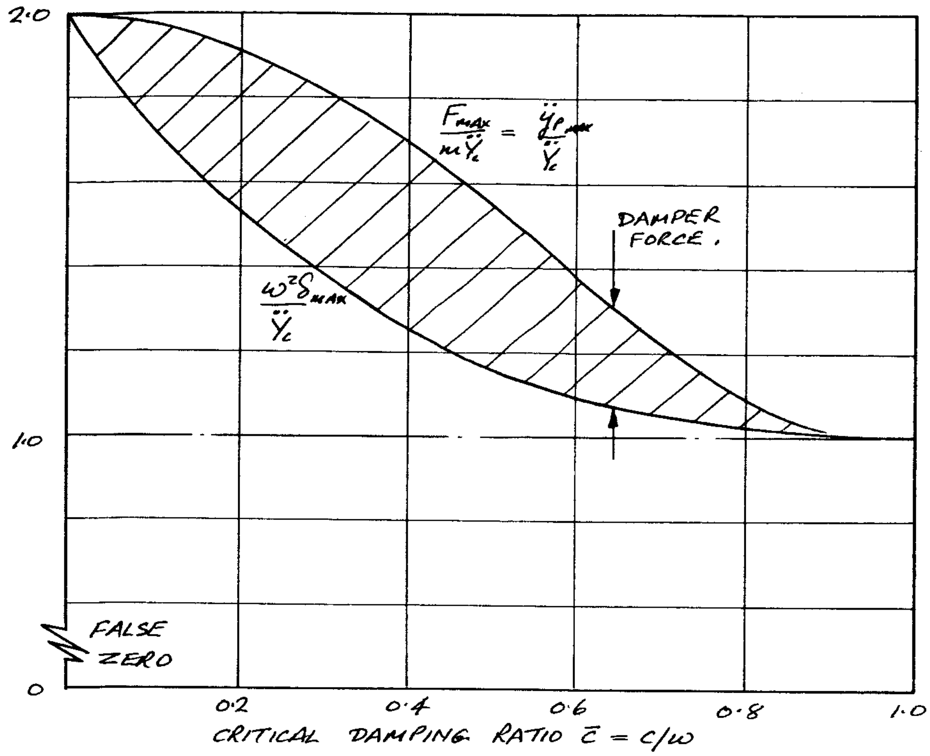


Figure 14. Peak Spring and Total Loads in a Damped System for Zero Rise Time Input

In order to study the effect of these differences upon physiological tolerance limits we have to derive the "spike input" relationship corresponding to equation 14. We do this by writing $\dot{\delta}_0 = \Delta v, \delta_0 = \ddot{Y}_c = 0,$ in equation 7.

i.e., $$\delta = \frac{\Delta v}{\lambda} e^{-\bar{c}\omega t} \sin \lambda t \qquad (15)$$

and $$\dot{\delta} = \Delta v e^{-\bar{c}\omega t} \{\cos \lambda t - \frac{\omega \bar{c}}{\lambda} \sin \lambda t\} \qquad (16)$$
$$= \frac{\Delta v}{\eta} e^{-\bar{c}\omega t} \sin \{ \lambda t + \sin^{-1}\eta \}$$

Thus δ is a maximum when

$$\lambda t + \sin^{-1}\eta = \eta\omega t + \sin^{-1}\eta = n\pi$$

i.e. when $\lambda t = n\pi - \sin^{-1}\eta$

and $\sin \lambda t = \eta$

$$\therefore \omega^2 \delta_{max} = \omega \Delta v e^{-\bar{c}\theta_m} \qquad (17)$$

where $\theta_m = \frac{1}{\eta}(\pi - \sin^{-1}\eta)$

The peak spring force given by equation 17 is compared with the peak total force in Figure 15. It is obvious that the two criteria give totally different results for high damping ratios, although the difference is unimportant for $\bar{c} < 0.15$, and a marked

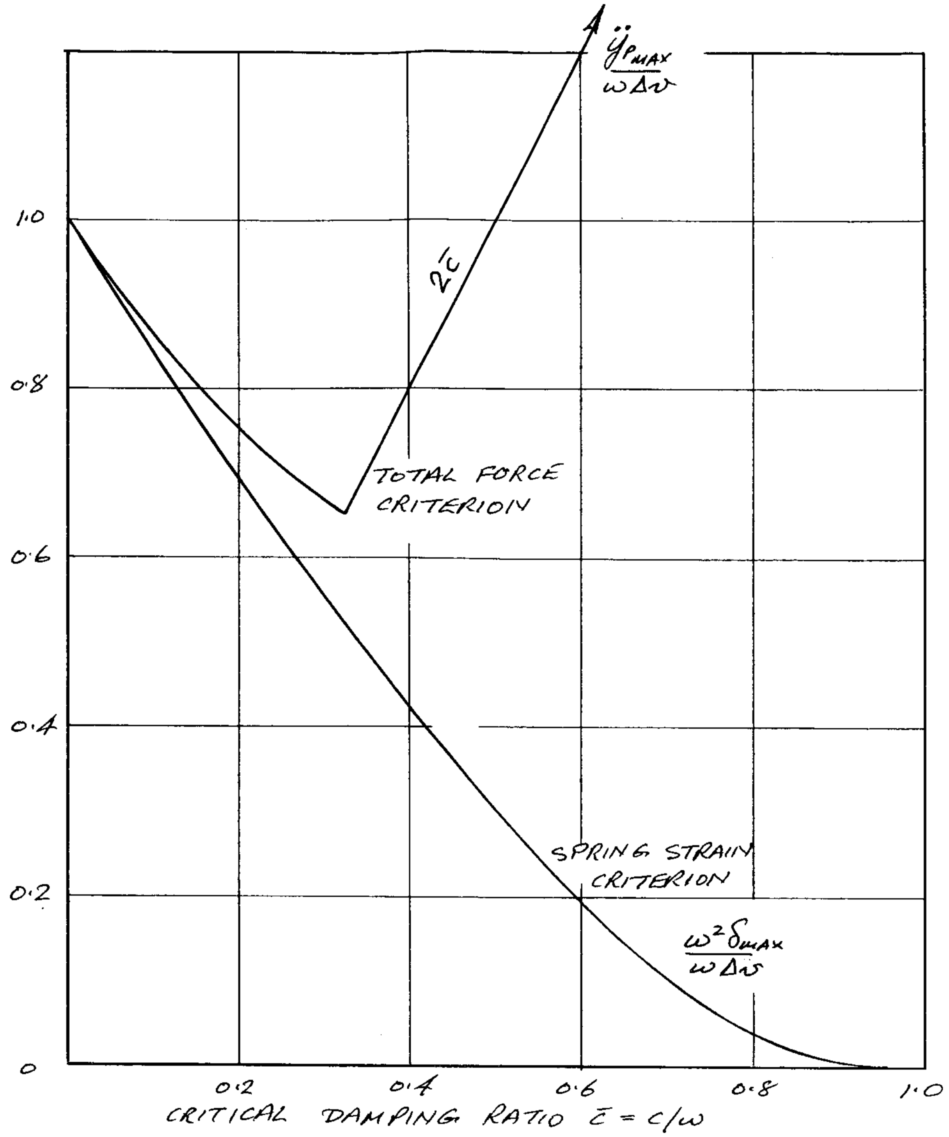


Figure 15. Peak Force and Spring Strain Criteria in the Impact Acceleration Range

divergence does not occur until we reach $\bar{c} = 0.33$, when the peak total force occurs at the moment of impact, and is entirely due to the damper.

We can now utilize the foregoing derivations to draw non-dimensional tolerance graphs, for both critical strain and critical total force assumptions. These are presented in Figures 16 and 17, and it is obvious that although they are very similar at low damping ratios, they become very dissimilar for the higher damping coefficients.

The experimental data currently available shows good agreement with the limit envelope for $\bar{c} = 0$, but it is evident that the same agreement would be obtained if we assumed damping of 10 per cent or even 20 per cent of the critical value. In many cases (such as analog computer work[1]) the inclusion of some damping is a distinct advantage. Although we can only say that the value of $\bar{c}$ should be less than 0.2, based on presently available data, we can expect to "zero-in" on a more specific number as

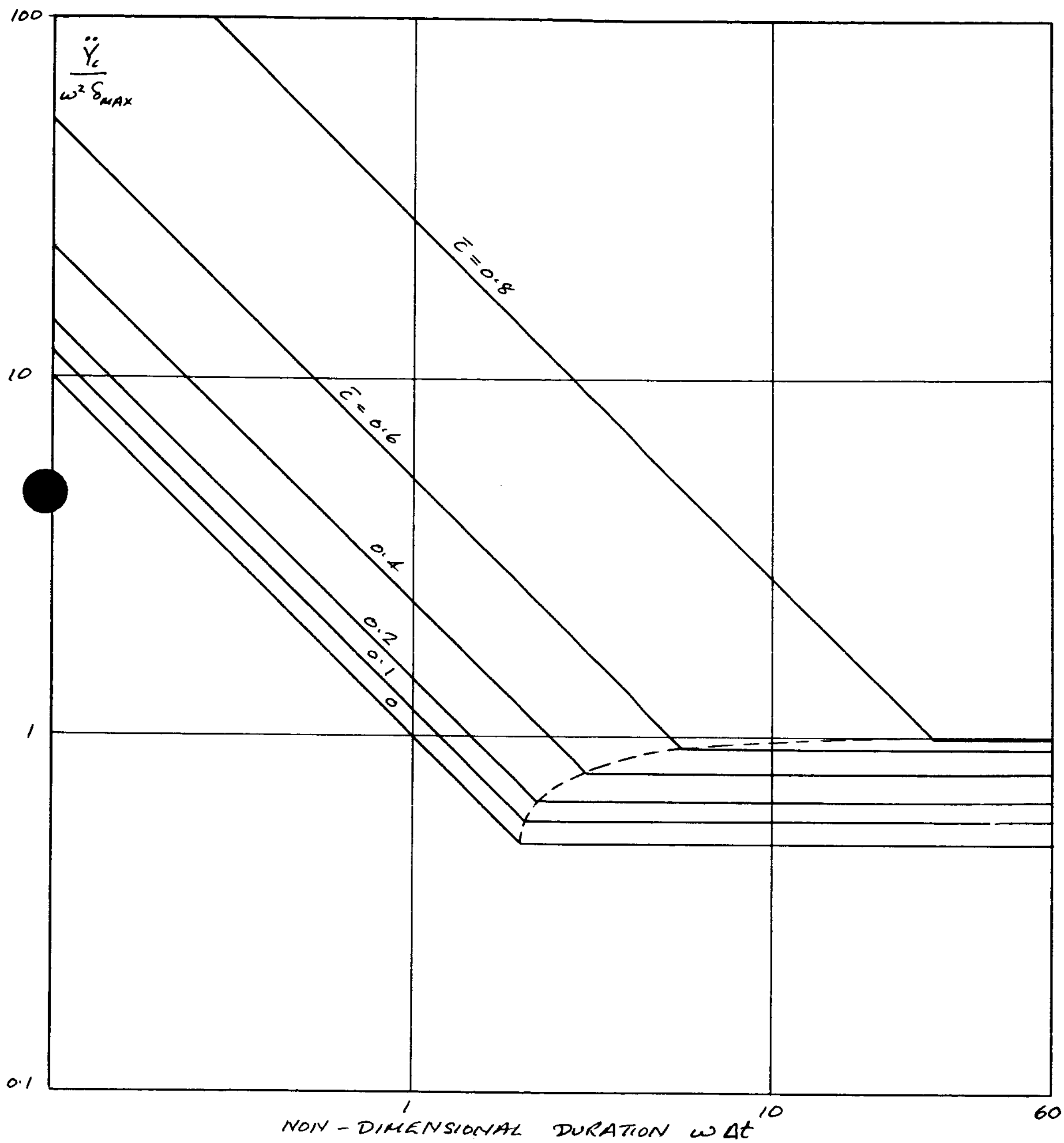


Figure 16. Non-dimensional Tolerance Envelopes Based on Critical Spring Strain Criterion, for a Damped System

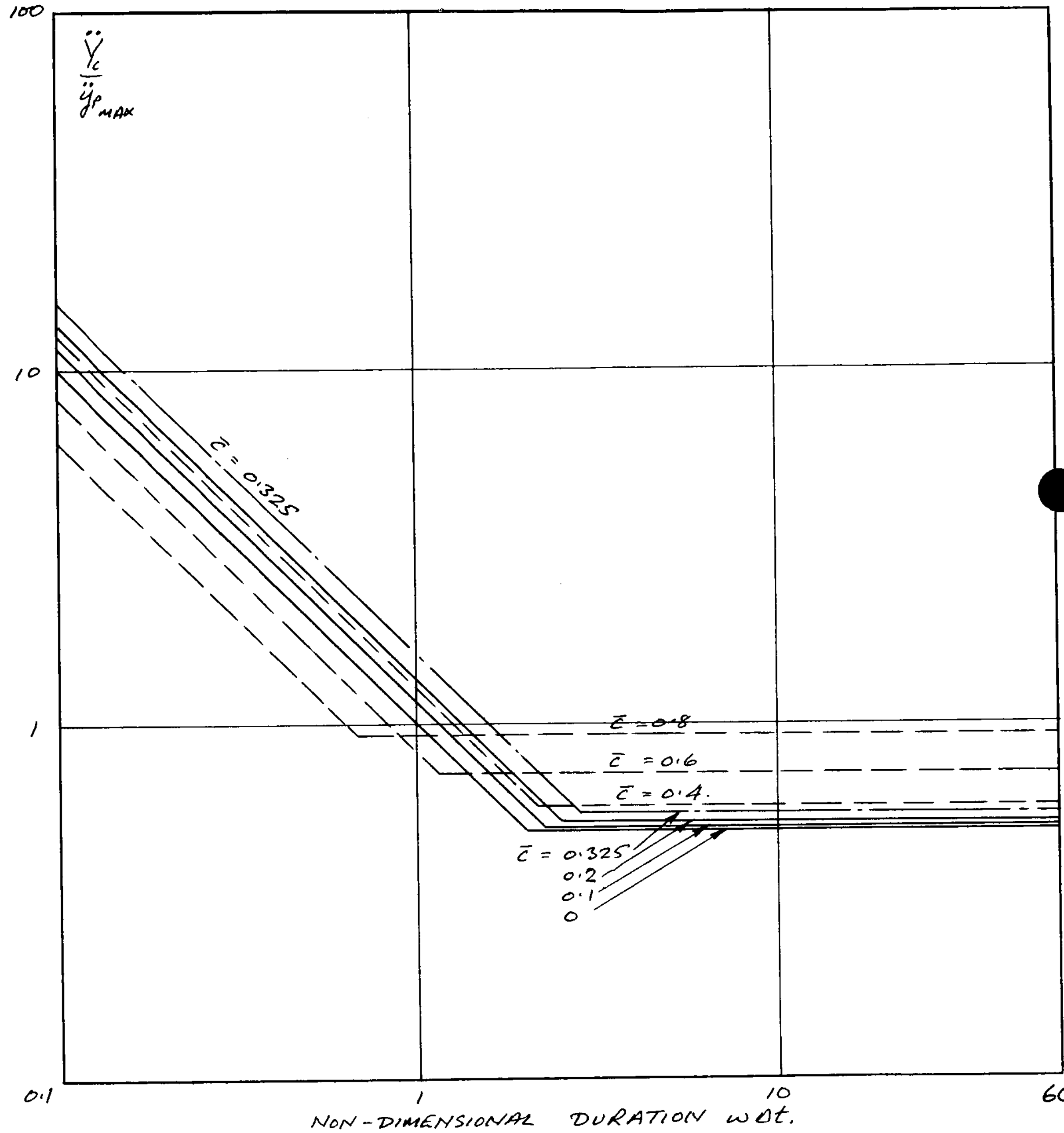
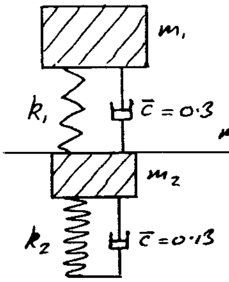


Figure 17. Non-dimensional Tolerance Envelopes Based on Critical Total Force Criterion, for a Damped System

more experimental data becomes available. The value currently used by the writer is $\bar{c} = 0.13$, which leads to the following models:

Spinal Model



"Viscera Mode"

ω_1 = 14.63 rads/sec.
= 2.33 cps
m_1/m_2 = 1.8

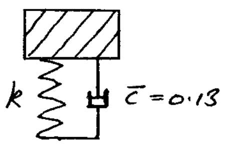
"Spinal Mode"

ω_2 = 287.5 rads/sec.
= 45.8 cps
δ_{CRIT} = 0.0258 ft.
= 0.31 ins.
$\omega_2^2\delta_{CRIT}$ = 66.2 g

The significant parameters derived from these results, for rectangular acceleration inputs are as follows:

	Viscera Mode	Spinal Mode
"Corner Duration" Δt_1	0.273	.0073 (secs)
Critical Velocity Change $(\Delta t < \Delta t_1)$	126.0	9.27 (ft/sec)
Critical short duration input $(\Delta t > \Delta t_1)$	23g	40g (ft/sec^2)

Transverse Model



ω = 49.6 rads/sec.
= 7.9 cps
δ_{CRIT} = 0.865 ft.
= 10.38 ins.
$\omega^2\delta_{CRIT}$ = 66.2g

The significant parameters derived from this model for rectangular acceleration inputs are as follows:

"Corner Duration" Δt_1	.0444 (secs)
Critical Velocity Change $(\Delta t < \Delta t_1)$	53.6 (ft/secs)
Critical long duration input $(\Delta t > \Delta t_1)$	40g (ft/sec^2)

Lateral Model

As for the zero damping models, the absence of experimental data forces us to use the transverse data for the lateral model.

The Effect of Non-linear Springs

It has been known for a considerable number of years that the "springs" of the human body are non-linear, in that their stiffness increases with deflection.

A convenient method of representing this non-linearity in a perfectly general way is to write

$$\text{Spring force} = \zeta_n \delta^n \tag{18}$$

where n can have any (constant) value, ζ_n is a constant.

Incorporating this express, the question of motion for the system of Figure 13 becomes

$$\ddot{\delta} + 2c\dot{\delta} + \zeta_n \delta^n = \ddot{y}_c \tag{19}$$

This equation has been solved(8) in the general case for zerp damping, and it can be shown that for any value of n the pulse acceleration tolerance graph is of the shape shown in Figure 18. Thus whatever value of n we select, we can always get good agreement with experimental data: there is, therefore, no advantage to be gained from using a non-linear system, and the remarks at the end of this Section are equally applicable to this case.

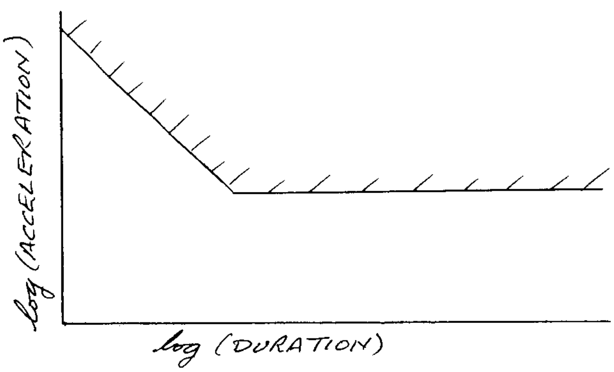


Figure 18. Non-linear Tolerance Graph for Pulse Acceleration

As physiological experiments become more refined, and particularly as the work at ASD Bio-Med Laboratory continues, it will become possible to assign a value to the non-linear exponent n with increasing accuracy. The ability to do so is extremely important to the successful application of body dynamics to capsule stability problems, to cite only one example, and it is to be expected that more attention will be focused upon this problem in the future.

It should be noted that non-linear theory is already well developed for the special case of personnel safety nets, due to a continuing research program at Frost Engineering which is funded by the Rose Manufacturing Company of Denver. This subject can be regarded as a limit case of restraint and is summarized briefly in the later Section on Safety Nets.

Models with Many Degrees of Freedom

As the nation-wide experimental program accelerates, we can expect a rapid increase in experimental information which will necessitate refinements in our dynamic models involving additional degrees of freedom. An example is provided by the model reported by Goldman and von Gierke in Reference 7, and illustrated in Figure 19. It is of course impossible to delineate the various stages of model evolution with any pretense of accuracy, since this will depend largely upon the overall emphasis of the experimental programs initiated in the future.

Hydraulic Tolerance Limits

It is generally agreed that for very long duration accelerations, the limits of human tolerance are a function of mass blood flow in a direction opposite to that of the acceleration vector. This phenomenon is susceptible to mathematical analysis,

particularly since the low blood velocities involved must mean that the system losses are essentially linear (viscous flow). In Figure 20 an element of fluid of tube area A and thickness dh has a force

$$\ddot{y}_c \rho A \, dh \tag{20}$$

acting upon it, due to the acceleration field. This is balanced by a velocity force

$$K_1 v \sqrt{4\pi A} \; dh \tag{21}$$

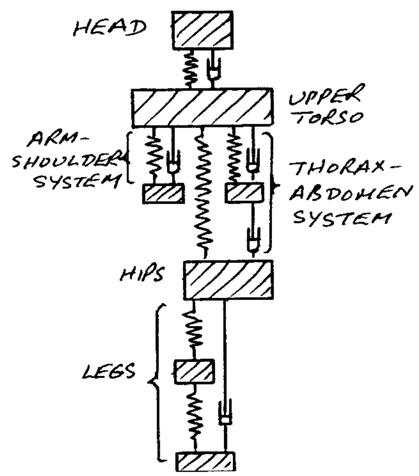


Figure 19. Mechanical Analog of the Human Body (Goldman and von Gierke)

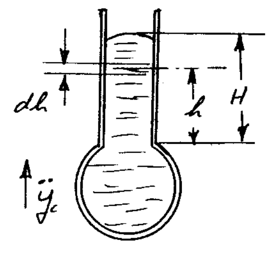


Figure 20. Hydraulic Analogy

If we neglect the elastic restraint of the blood vessels, and assume that only the action of the heart and viscous effects delay drainage of the blood to the "lower" parts of the body, then we obtain a very simple equation by equating Equation 20 to Equation 21.

i.e., $v = K_2 \ddot{y}_c$

mass flow $\dot{m} = v t \rho = K_2 \rho \ddot{y}_c t$ (22)

$= K_2 \rho \Delta v$ for constant accelerations

This result also neglects inertia terms associated with acceleration of the blood, which will limit its accuracy at high accelerations.

The elastic term can be taken as a linear function of mass flow $\dot{m}$

i.e., force $= K_3 \dot{m} = K_4 \int_0^t v \, dt$ (23)

Thus we now have as an equation of motion

$$\dot{\delta} + \omega^2 \delta = K_5 \ddot{y}_c$$

which gives $-\log\left[a \ddot{y}_c + \delta\right]_0^{\delta} = ct$ (24)

Thus the critical time is

$$t_{MAX} = \frac{1}{c} \log \left[\frac{a \ddot{y}_c}{a \ddot{y}_c - \delta_{CRIT}} \right] \tag{25}$$

The critical acceleration is

$$\ddot{y}_{CRIT} = \frac{(\delta/a)}{1 - e^{-ct}} \tag{26}$$

If ng is the tolerable acceleration for an infinite length of time, and $G_{CRIT} = \ddot{y}_{CRIT}/g$ then we can produce a set of curves of the type shown in Figure 22 for various values of C. Curve (A) from Figure 21 represents the practical case which nearest approaches the idealized conditions of this inertialess model, and the agreement is seen to be quite good.

For the other cases in Figure 21 it is evident that the neglected inertia terms will have to be included if we are to obtain a model which adequately describes the measured tolerance limits.

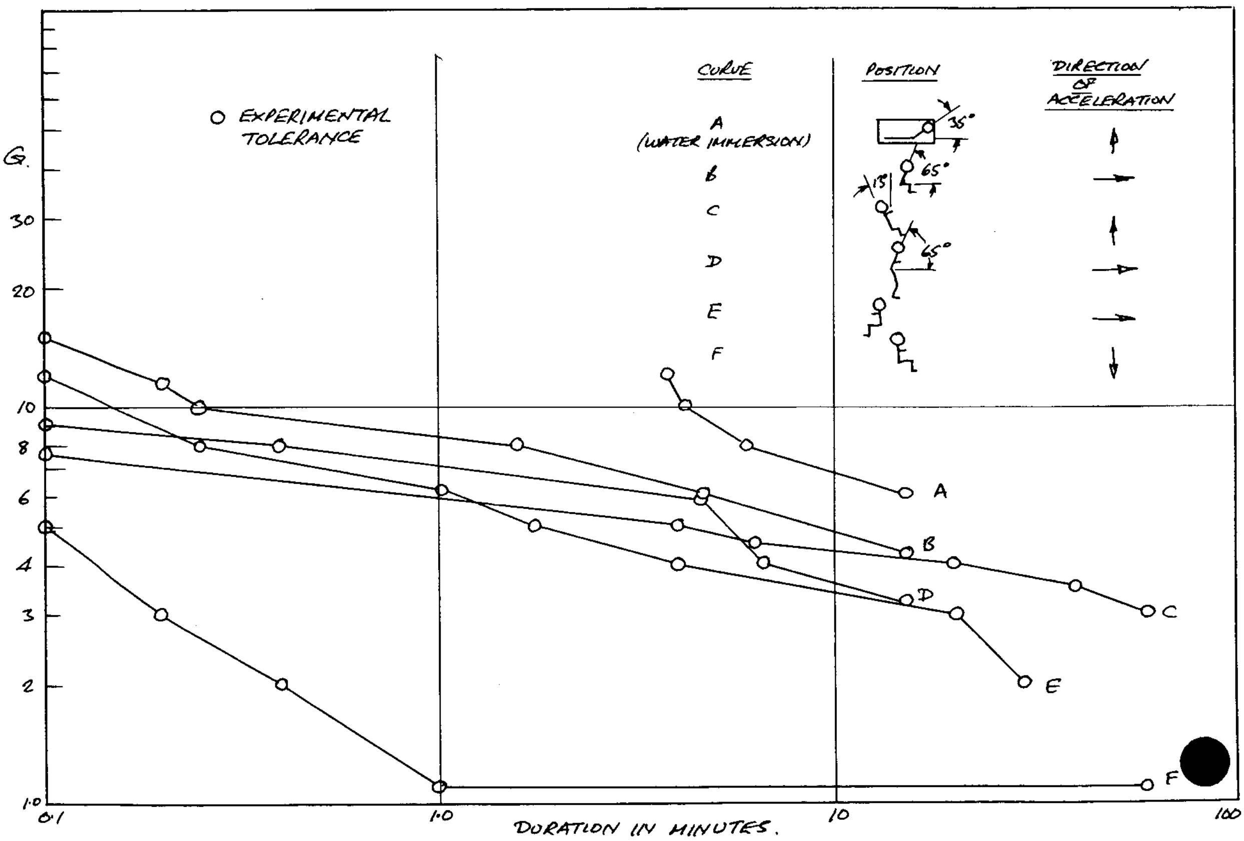


Figure 21. Tolerance to Long Period Acceleration (Ref. 16)

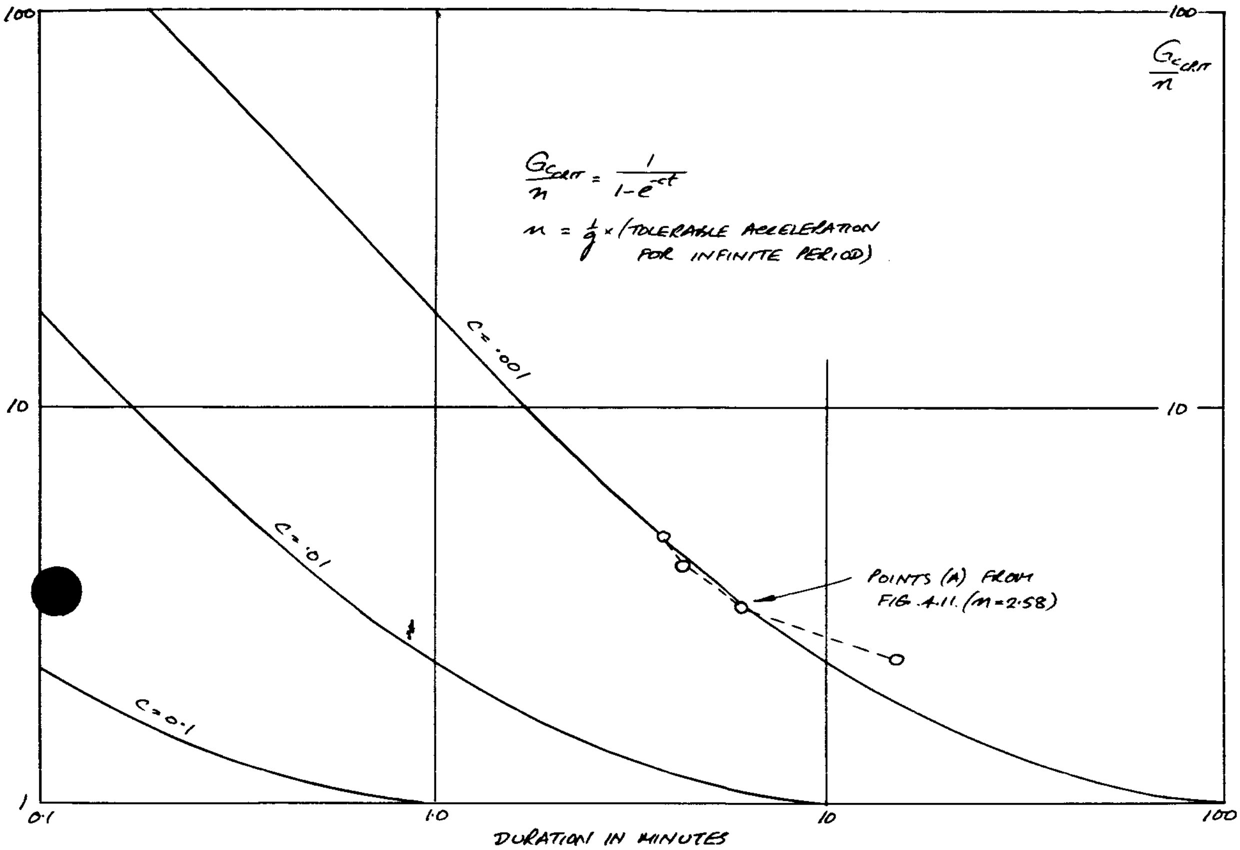


Figure 22. General Hydraulic Limit

It is already evident that the preceding equations for the hydraulic model of Figure 20 also apply to incomplete models of the familiar spring-mass system with damping. The inclusion of inertia terms now result in the conventional system of Figure 23 for which the equation of motion is

$$\ddot{\delta} + 2c\dot{\delta} + \omega^2\delta = \ddot{y}_c \tag{27}$$

Because hydraulic losses predominate, this system is supercritically damped, however, so that its motion will be dead beat. As before, we assume that the limit of human tolerance is defined by some critical spring strain δ_{CRIT}, which represents either a critical blood pressure or a critical blood-vessel stress.

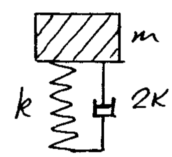


Figure 23

For zero initial conditions the supercritical solution to equation 27 is, for a long period acceleration of magnitude $\ddot{Y}_c$

$$\delta = \frac{\ddot{Y}_c}{\omega^2}\left\{1 - e^{-\frac{c}{\lambda}\cdot\lambda t}\left[\cosh\lambda t + \frac{c}{\lambda}\sinh\lambda t\right]\right\} \tag{28}$$

where $c = K/m \qquad \omega^2 = k/m$

$\lambda^2 = c^2 - \omega^2 = \omega^2(\bar{c}^2 - 1)$

$\bar{c} = c/\omega > 1{\cdot}0$

From our hypothesis, the critical acceleration magnitude as a function of the non-dimensional time λt is given by

$$f = \frac{\omega^2 \delta_{CRIT}}{\ddot{Y}_{c_{CRIT}}} = 1 - e^{-\frac{c}{\lambda}\lambda t}\left[\cosh \lambda t + \frac{c}{\lambda}\sinh \lambda t\right] \tag{29}$$

$$\frac{df}{dt} = \left\{\left(\frac{c}{\lambda}\right)^2 - 1\right\} e^{-\frac{c}{\lambda}\lambda t} \sinh \lambda t \tag{30}$$

For very long durations $\frac{\omega^2 \delta_{CRIT}}{\ddot{Y}_{c_{CRIT}}} \rightarrow 1.0$

Numerical solutions to equation 29 are plotted in Figure 24, and it is obvious from Figure 25, in which the reciprocals of the tolerance curves in Figure 21 plotted, that the experimental results are at least of the same class as the experimental limits. The determination of the damping and frequency values which give the best fit must wait until an appropriate computer program can be established, however.

It is hoped to obtain funding for a research program to establish models which adequately represent the curves given in Figure 21, by means of an appropriate computer program, and to integrate these models with the short period acceleration models established earlier. These models can then be used to establish the tolerability of any arbitrary acceleration-time history with a total duration of up to as long as one hour.

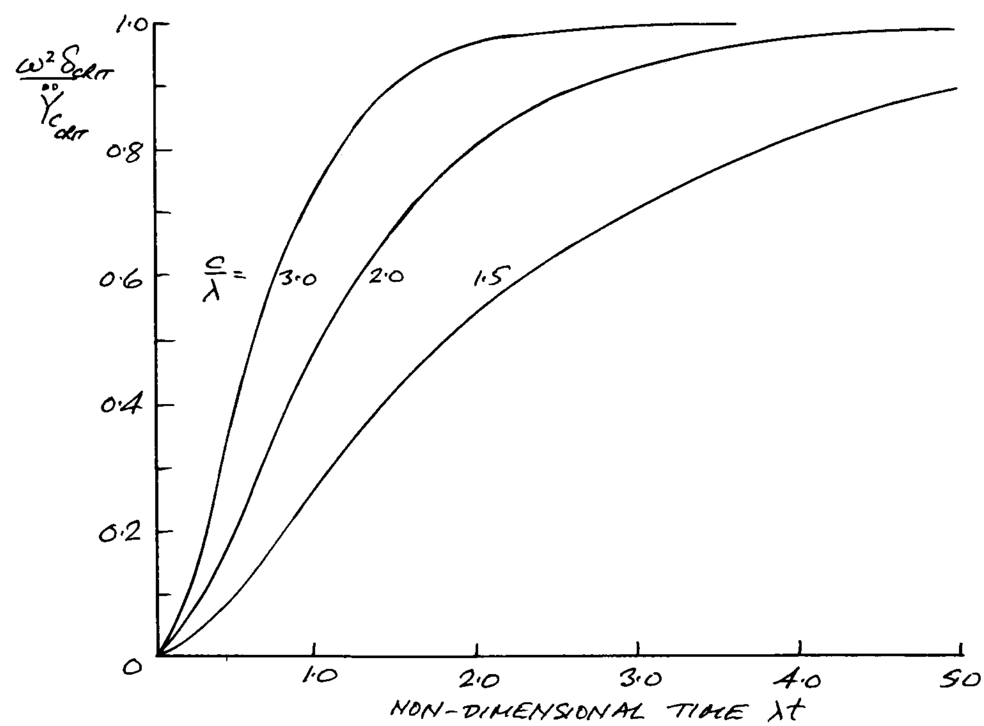


Figure 24. Numerical Values of Equation 22

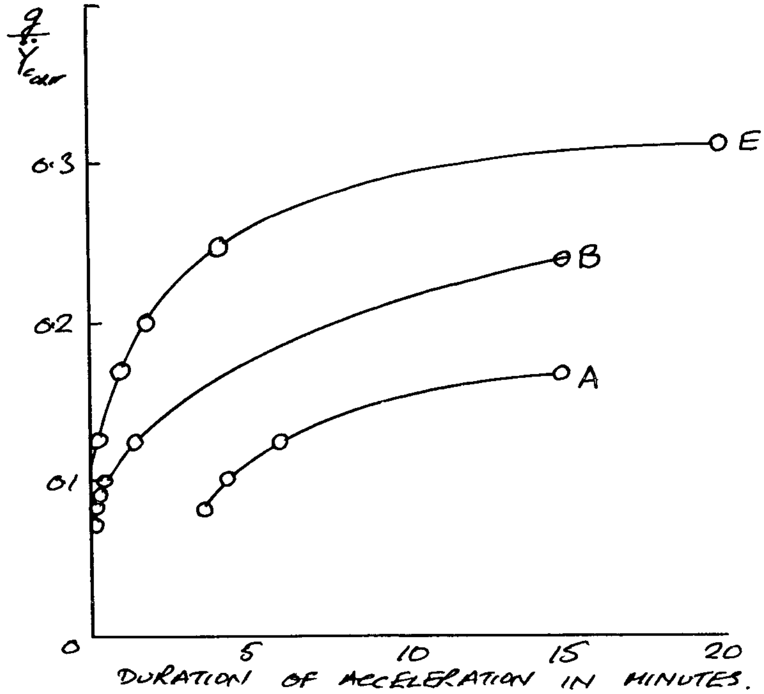


Figure 25. Reciprocal of Three Curves in Figure 21

The Dynamics of Head Restraint

It is common knowledge that a blow to the head can cause unconsciousness, or even death, and that this is due to the acceleration applied to the skull and hence transmitted to its contents.

Many cases arise where the acceleration applied to a human subject is not uniform, but varies along its length. For example, the "spinal ground landing system"(18) employed by the B-70 escape capsule can cause it to topple in a drift landing, even with its stabilizing booms, so that a secondary impact occurs when the capsule falls on its side or front, or could even result from the boom impact effects in a back landing. In the absence of bounce this gives a velocity change Δv variation of the type shown in Figure 26, the highest accelerations occurring at the position of the occupant's head. It is obvious that "gross body acceleration" data will be of little use in assessing the tolerability of such an impact, since although the accelerations in the area of the chest may be below the tolerable limit, the acceleration in the area of the subjects' head may be sufficient to render him unconscious.

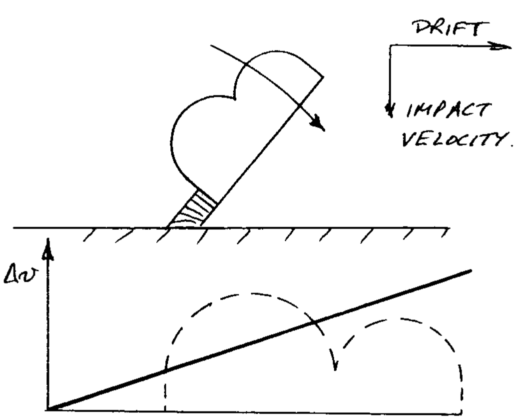


Figure 26. Transverse Velocity Change in Toppling Impact

Another familiar example of the need for defining tolerable head accelerations occurs when head impact or "neck whip" problems occur. What, for example, is the maximum velocity change which can be tolerated by a well supported head in an

"impact acceleration?" And what are the optimum dynamic characteristics of a flight helmet for a given level of impact protection?

Certain minimums can be established from "gross body" acceleration data. In the spinal direction we know that the limiting impact velocity change Δv must be greater than 9.3 ft/sec, and the critical long duration acceleration more than 40g, because these limits are defined by spinal damage rather than loss of consciousness.

Similarly, we know from experiments in which a transverse acceleration was applied that a well supported head can withstand at least 20 ft/sec in this direction. Since the onset of shock is generally the limiting factor in transverse accelerations, however, it is possible that this limitation is in fact due to some mechanism contained in the skull. Thus we currently have no justification for assuming that higher head accelerations are safe.

To construct a dynamic model it is not necessary to know the mechanism which causes an end-point to be reached and any physically reasonable hypothesis which gives agreement with the available experimental data can be used. In the present case, however, there is no readily available data, so that the discussion must be confined to the different classes of dynamic models which can be used.

In Reference 7, Goldman and von Gierke state:

" Many investigators consider .... shear strains resulting from rotational accelerations due to a blow to the unsupported head as the principal event leading to concussion. Blows to the supported, fixed head are supposed to produce concussion by compression of the skull and elevation of cerebrospinal fluid pressure."

Both of these mechanisms are susceptible to mathematical analysis.

A Pressure Model of the Head

If ℓ is some characteristic length in the direction of acceleration, then the maximum pressure in the head, caused by the acceleration will be

$$p_{MAX} = \ddot{y}_c \rho \ell \tag{31}$$

If the acceleration is applied rapidly (very small rise time) then we know from classical mechanics that the pressure generated will be twice this value.

The equation of motion governing the propagation of pressure waves is the familiar wave equation

$$\frac{\partial^2 \xi}{\partial x^2} = \frac{1}{c^2} \frac{\partial^2 \xi}{\partial t^2} \tag{32}$$

where ξ = strain
x = lengthwise position
c = velocity of sound in the medium

The pressure generated is

$$p = E \frac{d\xi}{dx} \tag{33}$$

where E = modulus of elasticity
$d\xi/dx$ = strain per unit length

The general solution to equation 32 is

$$\xi = (A \sin kx + B \cos kx)(C \sin \omega t + D \cos \omega t) \tag{34}$$

where $k = \omega/c$

$A - - - D$ are arbitrary constants of integration

It can be shown[14] that the behavior of this system is very similar to that of a spring mass system.

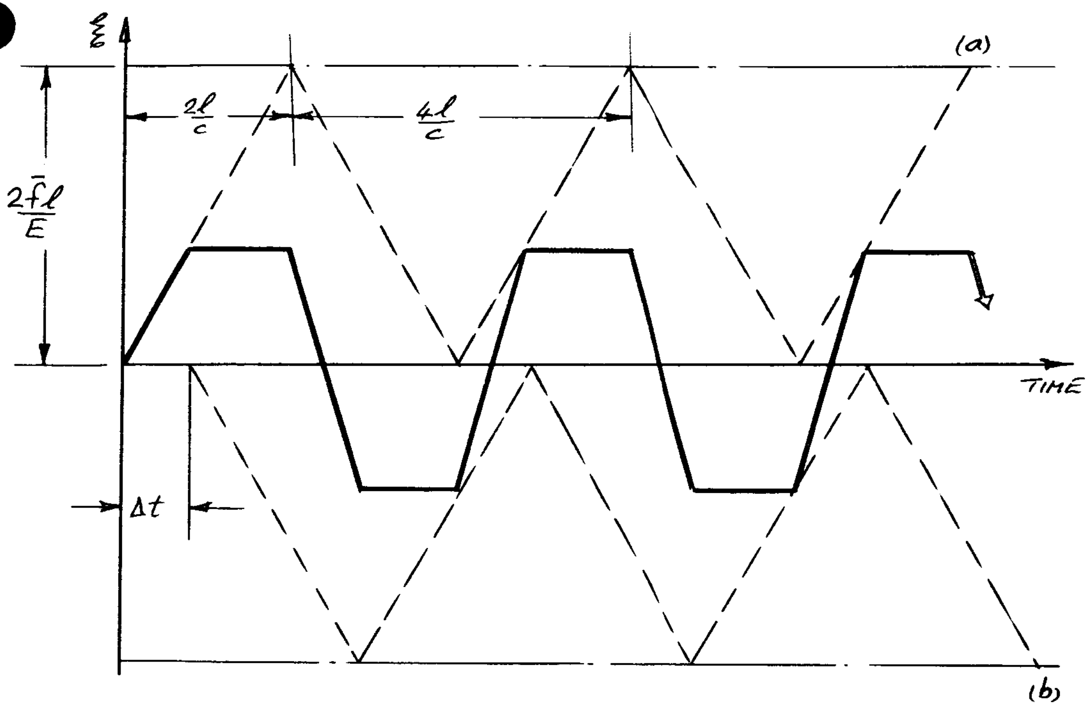


Figure 27. Stress-response of Homogenous Bar to a Pulsed Force of Duration Δt Seconds

From Figure 27 the response is

$$\xi_{MAX} = \frac{2\bar{f}\ell}{E}\frac{\Delta t\, c}{2\ell} = \frac{\bar{f}}{E} c\, \Delta t \qquad (\Delta t < 2\ell/c) \tag{35}$$

$$= \frac{2\bar{f}\ell}{E} \qquad (\Delta t > 2\ell/c) \tag{36}$$

where $\bar{f}$ = force per unit area

If m is the bar mass, and A its area, the gross body equation of motion will be

$$m\ddot{y} = A\bar{f}$$

whence $$\dot{y} = \frac{A}{m}\bar{f}t \qquad (37)$$

and the velocity change associated with a pulsed force of duration Δt will be

$$\Delta v = \frac{A}{m}\bar{f}\Delta t = \bar{f}\Delta t/\rho_B \ell$$

whence $$\bar{f} = \frac{\Delta v \rho_B \ell}{\Delta t} \qquad (38)$$

From the equations 35 and 36 the impact stress or pressure in the material is

$$p_{MAX} = \frac{\bar{f} c \Delta t}{\ell} = c \rho_B \Delta v \qquad (\Delta t < 2\ell/c) \qquad (39)$$

while the long duration pressure is

$$p_{MAX} = 2\ddot{y}_c \rho \ell \qquad (\Delta t > 2\ell/c) \qquad (40)$$

which agrees with the result of equation 31 when the rise time is zero.

Reference 7 gives the measured Young's modulus for a skull as 1.4×10^{10} dynes/cm^2 (2.03×10^5 lb/in^2)

giving $c = \sqrt{\frac{E}{\rho}} =$ 3,900 ft/sec

Taking ℓ to be 0.6 ft., the theoretical corner duration is therefore

$$\Delta t_1 = \frac{2\ell}{c} = \frac{1.2}{3900} = .000308 \text{ seconds}$$

We can now sketch a tentative tolerance graph for the human skull, based on the assumption that existing gross body transverse limits are imposed by brain injury in the long duration area, and that for duration of less than 0.4 seconds, the soft restraint of the head prevents transmission of the high "impact" accelerations (Figure 28). This is physically reasonable because most of the test points upon which the transverse limits are based were "eyeballs out," so that the head was not rigidly restrained.

There is some experimental evidence to suggest that the critical velocity change for a well supported head ("eyeballs in") is about 10-20 ft/sec. In this case, the maintenance of 40g would require a head movement of the order of six inches in an "eyeballs out" experiment. However, this invalidates the calculated "corner duration" of .00031 seconds, since the value for $\Delta v = 10$ ft/sec would be nearly .008 seconds, or alternatively, the effective modulus of elasticity of the head would have to be much lower than the value used in the above calculations.

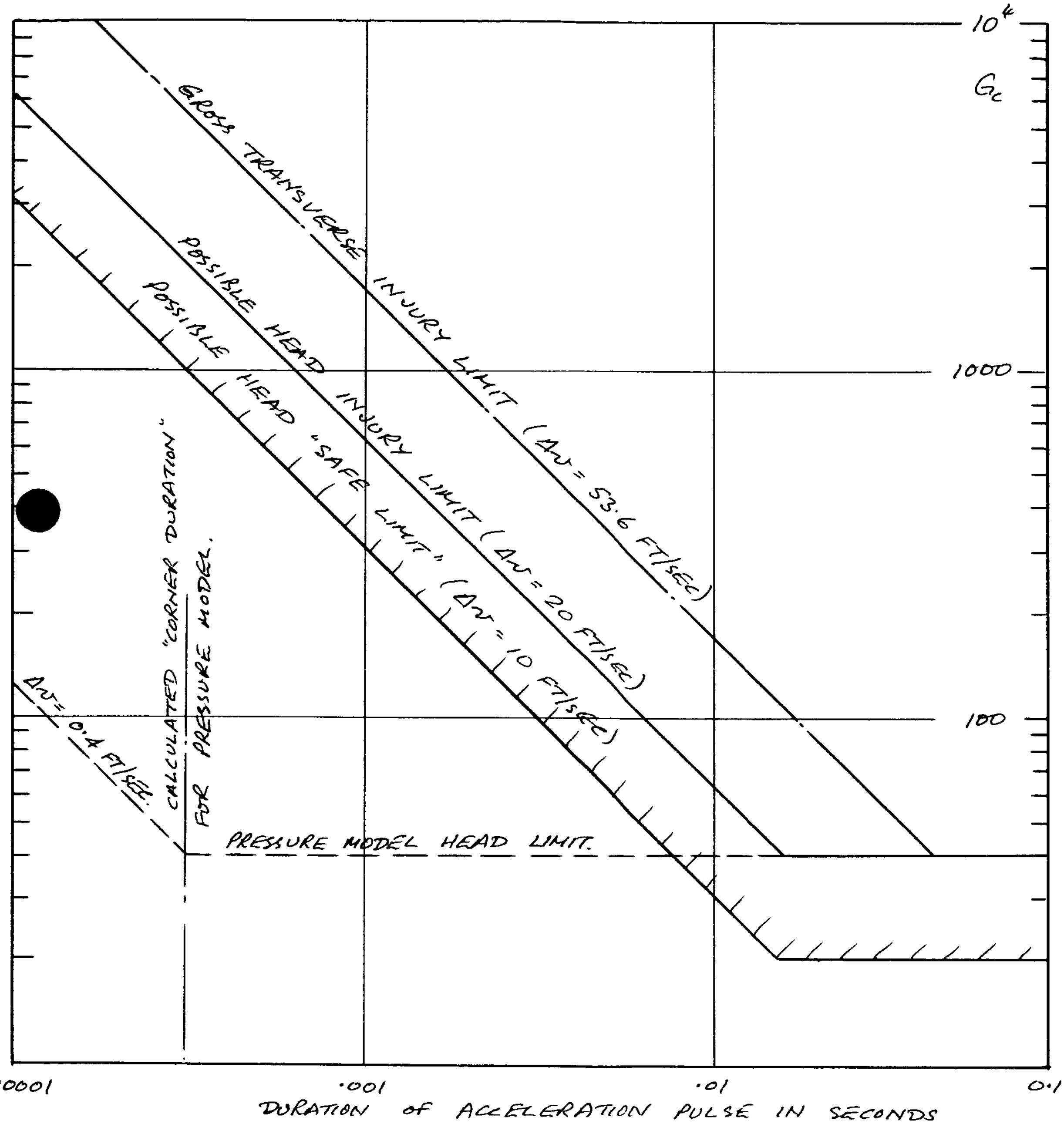


Figure 28. Head Acceleration Tolerance Limits

Nevertheless, the pressure model is of considerable interest because:

a. It gives some indication of the mechanics of pressure wave transmission.

b. It shows that the simple spring-mass system can be used for assessing head accelerations, because the equations for both are identical for practical applications.

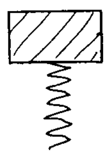
c. It indicates that the head is less able to withstand impacts because it is more rigid than the rest of the body.

d. Following (c), it points the way to the determination of optimum head protection helmets.

A Mechanical Model of the Head

Since the pressure model gives the same equations for practical purposes, as an equivalent spring-mass system, it is convenient to use the latter because of the well developed state of the theory associated with it. Since this theory is discussed in the two previous sections, there is no need to repeat it here.

We tentatively assume that the $\Delta v =$ 20 ft/sec line in Figure 28 defines the limit of tolerability for the head in all directions, and that 40g is the "long duration" limit. This leads to the following model:

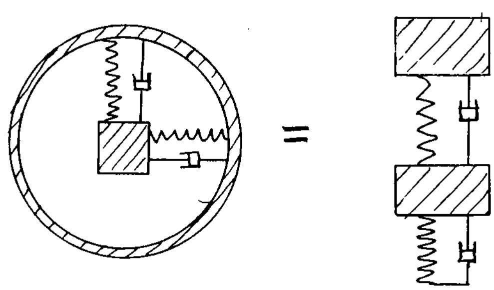


ω = 129.0 rads/sec
= 20.55 cycles/sec
δ_{CRIT} = 0.155 ft.
= 1.857 in.

"Corner duration" $\Delta t_1 = 2/\omega =$ .0155 secs
Critical velocity change = 20.0 ft/sec
Critical long duration input = 40g

The "safe tolerance limits" are obtained by halving the allowables calculated above.

When more precise data relating to head acceleration tolerances becomes available it may well prove that the response characteristics of the critical spring strain vary with the direction of application. It is also possible that we shall have to separate the skull and its contents into two degrees of freedom, so that the model will resemble Figure 29.



(in each degree of freedom)

Figure 29. Possible Development of Head Model

Head Restraint Dynamics

The principles and effects of head restraint systems are now seen to be identical with general restraint system dynamics, and the same methods can be used for calculating optimum restraint characteristics.

Restraint System Dynamics

A restraint system has three functions to perform; it must be comfortable under normal conditions and yet fully restrain its occupant under the highest expected g loading, and it must attenuate rather than magnify accelerations which are imposed by a vehicle upon its occupant(s).

In this paper we are primarily concerned with acceleration transmission, assuming that the restraint system loads are adequately distributed over the body, so that damage is not caused by high strap pressure or local dislocation of the body's structure. Restraint comfort is mainly a function of position and pressure loading, and is most efficiently accomplished by "cut and try" development once the optimum dynamic characteristics have been calculated and verified.

Acceleration transmission is a function of the resiliences between the vehicle's structure and the occupant, their non-linearities, and whether or not they "bottom-out."

These resiliences are due to:

a. Cushions (or net support)
b. Restraint harness
c. Flexibilities between the seat pan and the "rigid" structure
d. As a "limit" case, "crushable foam" or honeycomb

These effects can all be treated by linear theory (albeit with provision for discontinuities such as "cushion bottoming") or by the use of non-linear functions which approximate the damping and force-time history of real materials.

Preliminary dynamic investigations have already been carried out for some important linear cases, such as that depicted in Figure 30 when either the cushion stiffness or its damping is absent. The results appear to be in excellent general agreement with experiments, such as those carried out on the ACEL ejection tower(21). For example, a typical spring cushion will alter tolerance to a rectangular input in the manner shown in Figure 31. That is, the cushion will be beneficial for very short period accelerations in the "impact" range, but will magnify a longer period input.

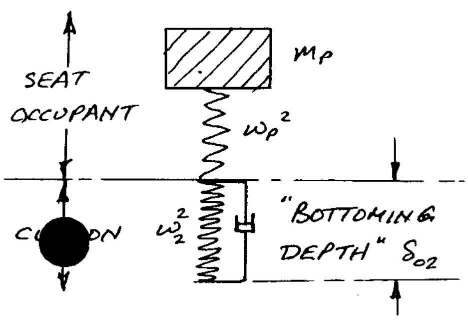


Figure 30

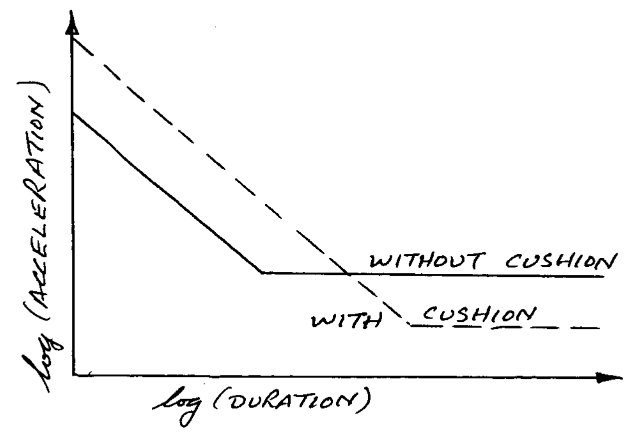


Figure 31

Thus an appropriate dynamic analysis is an essential preliminary to the intelligent use of restraint systems. It is certainly not possible to assume that, because a certain material and restraint geometry were beneficial in a certain series of tests, the same system will be equally beneficial for any other type of acceleration-time input.

Linear Restraint System Which Does Not Bottom

A restraint harness does not usually bottom out, although the non-linearity of its load against extension characteristics may result in its effective linear spring rate increasing with load. The same is true of a net seat or couch, provided that sufficient space exists around it to permit adequate stretch.

In an impact, if we consider only the linear case, without damping, then from Figure 30, the critical velocity change is

$$\Delta v_{CRIT} = \frac{\ddot{y}_{CRIT}}{\omega} = \frac{\ddot{y}_{CRIT}}{\omega_p \omega_2}\sqrt{\omega_p^2 + \omega_2^2} \qquad (41)$$

But $\omega_2^2 = k_2 / m_p$

and $$\ddot{y}_{CRIT} = \frac{k_2 \delta_{CRIT}}{m_p} = \omega_2^2 \delta_{CRIT} \qquad (42)$$

and therefore the maximum deflection is

$$\delta_{CRIT} = \frac{\Delta v^2}{\ddot{y}_{CRIT}} - \frac{\ddot{y}_{CRIT}}{\omega_p^2} \qquad (43)$$

For the very short period spinal case, $\omega_p^2 = (278)^2$ $\delta_{CRIT} = .0333$ ft. so that we have

Δv	=	10	20	30	40	50	ft/sec.
δ_{CRIT}	=	.061	.294	.682	1.225	1.92	ft.
	=	0.72	3.53	8.18	14.7	23.05	inches

With a required maximum deflection of only 8-inches to attenuate an impact of 30 ft/sec it is obvious that a net seat could be a very practical solution, particularly since almost no attenuation would be required in the transverse and lateral directions. However, it should be noted that the practical deflection would be larger than 8-inches, because of the non-linear behavior of net. The non-linear theory developed for safety nets (see later section) is applicable to this problem.

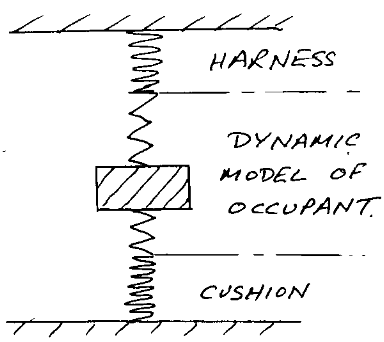


Figure 32

It must be emphasized here that the couch cushion and the restraint harness cannot be considered as separate items, since in the absence of damping, the cushion net will throw the occupant out again and into the harness, with a velocity equal to that of the impact. The appropriate model of an integrated harness-cushion system can be sketched as in Figure 32, so long as it is remembered that the motion from cushion to harness, and vice-versa is discontinuous. It is then easy to show that, under any type of acceleration:

$$\frac{\text{Rebound acceleration in harness}}{\text{Initial peak acceleration on cushion}} = \frac{\omega_1}{\omega_2} \qquad (44)$$

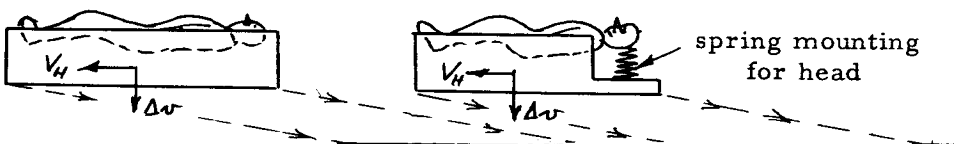
where ω_1 = natural frequency of occupant on the harness
ω_2 = natural frequency of occupant on the cushion

Thus there exists for any particular design of landing system and capsule occupant restraint, a maximum critical stiffness for the restraint harness, above which it will impose intolerable accelerations on the rebound cycle. In some practical cases this critical stiffness will be lower than the theoretical value, because of the localized loads which it applies to the body structure.

Rigid Contoured Support

The use of a rigid support which is contoured to fit the human body closely has been suggested by a number of investigators, such as Kornhauser[9, 10] for example, who suggested that such a method would be suitable for impacts up to 80 ft/sec.

The methods used in this report suggest that the true performance of a rigid couch dropped on concrete is much more complicated than would be indicated by the simple considerations used by Kornhauser, and that the vertical allowables would depend upon whether the subject's head is shock mounted or directly in contact with the rigid couch, since from Figure 28, the impact tolerance of a head is thought to be substantially lower than for the rest of the human body.



	Safe	Limit	Safe	Limit	Kornhauser[9, 10]	
Vertical velocity	10.0*	20.0*	26.8	53.6	80.0	/sec
Spinal velocity	4.7	9.3	4.7	9.3	-	
Lateral velocity	10.0*	20.0*	26.8	53.6	-	

* Limited by head acceleration tolerance

Figure 33. Performance of a Rigid Contoured Couch on Concrete

These tolerance limits can be significantly greater for a water landing, particularly if a penetrating "hydro-ski" is deployed below the vehicle to cushion the initial impact shock.

It should be noted from Figure 33 that the tolerable spinal drift velocity change is very low for a rigid couch, so that although it may seem a satisfactory device for a purely vertical landing, it cannot be employed in a practical vehicle which usually has some wind drift velocity component to dissipate. Therefore, some form of shock attenuation must be inserted between the rigid couch and the ground.

If μ is the friction coefficient of the impact surface on hard ground and G_s the deceleration level imposed by the shock absorbing device, then the worst spinal deceleration which will be imposed is (μG_s) g's. Thus to stay within the tolerance limits of Figure 11, we have the condition

$$\mu G_s < 20.0 \qquad (45)$$

The use of such a low deceleration means that we are well into the "long duration" region for most practical cases, so that for $\mu < 1.0$ the limitation has now been transferred to the transverse mode. Thus shock absorbers which have a short acceleration rise time should not impose more than 20g's and long rise time deceleration should be limited to 40g's.

Contoured, Crushable Support

If a couch or seat cushion is made of crushable foam or honeycomb, then to a first order of accuracy we can assume that it will crush at a constant load R and will remain "rigid" for lower loads.

During the crushing process a finite couch mass $\Delta \bar{m}$ is also being accelerated, by an instantaneous force $\Delta \bar{m} \ddot{\Delta}$, where

Δ = deflection of crushable material
$\bar{m}$ = inertia of material per unit depth

Thus the total reaction of the couch is

$$R + \Delta \bar{m} \ddot{\Delta} = k S \tag{46}$$

It should also be noted that in this problem, some of the mass associated with the human body must be assumed to move with the couch so that the real effective inertia term is ($\Delta \bar{m}$ + some constant).

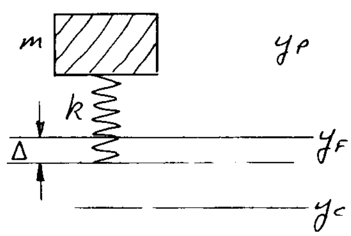


Figure 34

The crushable support problem is therefore one of two degree of freedom, and the analysis is considerably more complicated than would appear at first sight. But for the case where the mass of the crushable material is negligible, we have the elementary results

$$\left.\begin{aligned} \delta_{MAX} &= R/k \\ \text{or} \quad \ddot{y}_P &= R/m \end{aligned}\right\} \tag{47}$$

Linear Spring Support Which Bottoms

We can represent the load vs deflection characteristics of a real restraint system component, such as a cushion, by the linear approximation shown in Figure 35, with a discontinuity at the deflection δ_{o2}.

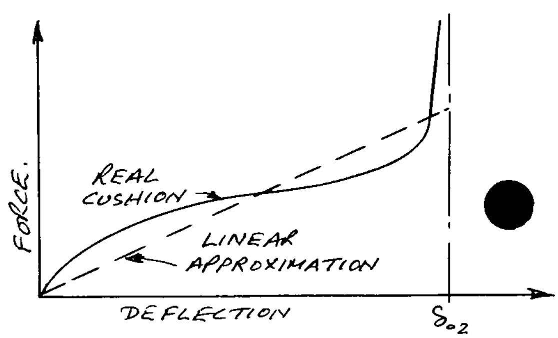


Figure 35. Linear Approximation for a Bottoming Cushion

We have already seen that the insertion of a purely "spring" cushion between the man and the structure which imposes an acceleration upon him has the effect of lowering his frequency, the relationship being

$$\text{effective } \omega = \omega_1 / \sqrt{1+\phi} \tag{48}$$

where $\phi = \omega_1^2/\omega_2^2 = k_1/k_2$,

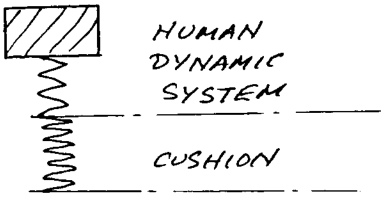


Figure 36. Simple Spring Representation of Cushion

The effect of this on the allowables is simple if we consider only the case in which the static acceleration $\ddot{y}_s$ is zero. For the impact range of accelerations

$$\omega_1 \delta_{1_{MAX}} = \Delta v / \sqrt{1+\phi} \tag{49}$$

and the corner duration Δt_1, which limits the applicability of velocity damage theory is

$$\Delta t_1 = \frac{2}{\omega_1} \sqrt{1+\phi} \tag{50}$$

From the equation 49, the softer the cushion, the less will be the physiological effect of impact acceleration, as shown in Figure 37.

ϕ	$1/\sqrt{1+\phi}$
0	1.000
0.1	0.954
0.2	0.913
0.3	0.878
0.5	0.817
1.0	0.707
2.0	0.578
3.0	0.500
8.0	0.333

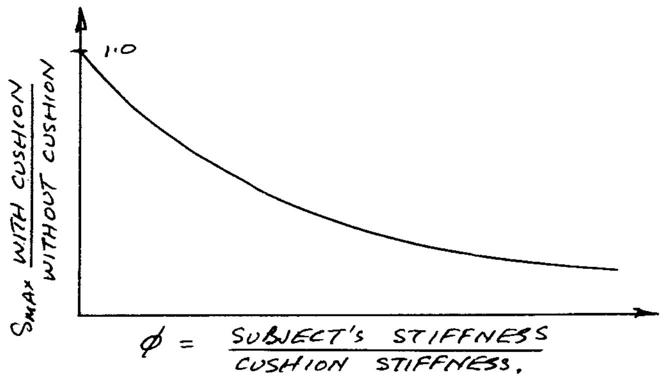


Figure 37. Effect of Cushion on Impact Forces

Another method of illustrating the effect of a non-bottoming resiliency is the acceleration tolerance plot of Figure 38, where the softer the restraint stiffness, the higher is the velocity change which can be tolerated by its occupant in the impact acceleration range. For long duration accelerations the cushion has no effect, since the factors governing 100 per cent overshoot still apply.

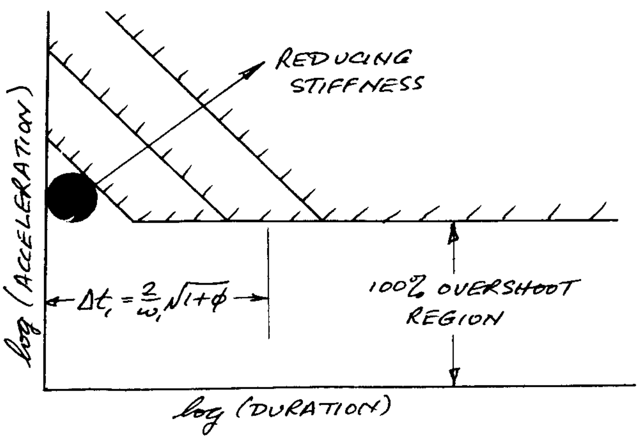


Figure 38. Effect of a Non-Bottoming Cushion on Acceleration Tolerance

In practical applications a cushion (for example) nearly always bottoms in the course of an impact, so that the equations are complicated by the presence of a discontinuity. A convenient way of treating this problem, for the impact case at least, is to use the "energy storage" parameter E_c ,which is defined as the total energy which a resiliency can absorb before bottoming. For a linear spring

$$E_c = \tfrac{1}{2} k_2 \delta_{02}^2 \tag{51}$$

$= \frac{1}{2}$ x stiffness x bottoming depth

For the general case of a non-linear cushion, E_c is defined as the work necessary to bottom it. A further parameter E_c' is the work necessary to bottom a cushion from some initial condition; usually the $1g$ condition. For zero initial conditions the strain in the human body model can be shown to be

$$\delta_{MAX} = \frac{\Delta v}{\omega_1}\sqrt{1 - \frac{2E_c}{m\Delta v^2}} \tag{52}$$

instead of $\Delta v/\omega_1$, for the man without a cushion. This result is perfectly general in the impact region, so long as the cushion actually bottoms. The velocity change necessary to cause bottoming is, for zero initial conditions

$$\Delta v_B = \sqrt{\frac{2E_c}{m}\frac{(1+\phi)}{\phi}} \tag{53}$$

Substituting Δv_B for Δv in equation 52, we obtain

$$\delta_{MAX} = \Delta v_1/\omega_1\sqrt{1+\phi}$$

which is the same result in equation 49.

The velocity change Δv_B required to just cause bottoming varies directly as $\sqrt{E_c/m}$. Its variation with stiffness ratio ϕ is indicated in Figure 39.

ϕ	$\sqrt{\frac{1+\phi}{\phi}}$
0	00
0.1	3.32
0.2	2.45
0.3	2.09
0.5	1.73
1.0	1.41
2.0	1.23
3.0	1.15
8.0	1.06

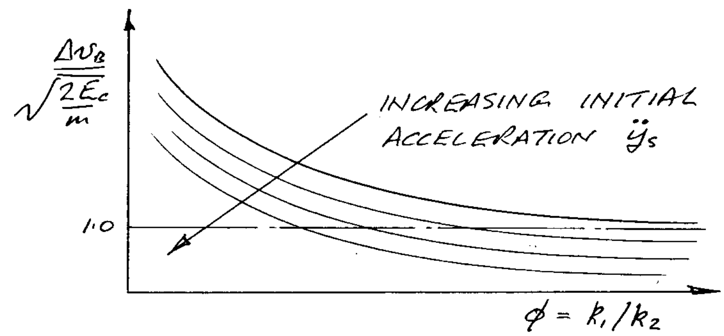


Figure 39. Variation of Bottoming Velocity Δv_B with Cushion Stiffness

It should be noted that the cushion stiffness has no physiological effect in an impact acceleration when bottoming occurs, but only the "energy storage" parameter E_c.

For long duration acceleration input $\ddot{Y}_c$ we need to introduce the concept of bottoming strain δ_B, which is the deflection of the human spring required to bottom cushion.

Since $\delta_1 k_1 = \delta_2 k_2$

$$\delta_B = \delta_{o2}/\phi \tag{54}$$

For zero initial conditions it can then be shown that the peak strain in the human spring is

$$\delta_{MAX} = \frac{\ddot{Y}_c}{\omega_1^2}\left\{1+\sqrt{1+2\left(\frac{\omega_1^2\delta_B}{\phi\ddot{Y}_c}\right)-(1+\phi-\phi^2)\left(\frac{\omega_1^2\delta_B}{\phi\ddot{Y}_c}\right)^2}\right\} \qquad (55)$$

This function is plotted in Figure 41, and it is obvious that when the cushion is softer than the human spring ($\phi > 1.0$) the worst possible case is when the cushion depth

$$\delta_{0Z} = \frac{\ddot{Y}_c}{\omega_2^2} = \frac{\ddot{Y}_c m}{k_2} \qquad (56)$$

i.e., when the static acceleration required to bottom the cushion is equal to the dynamic input acceleration $\ddot{Y}_c$. The variation of acceleration magnification with cushion stiffness is plotted in Figure 42, for this "worst case." The zero stiffness case with which this result is compared was obtained by calculating the acceleration due to the base of the human model being placed at a height δ_{02} above the accelerating surface at time t

Wholly Viscous Restraint

This system is an idealization of a cushion or other restraing component whose damping or dissipative characteristics are much greater than its ability to react a static load. Because it is a fairly academic concept, only the non-bottoming case will be described in this report.

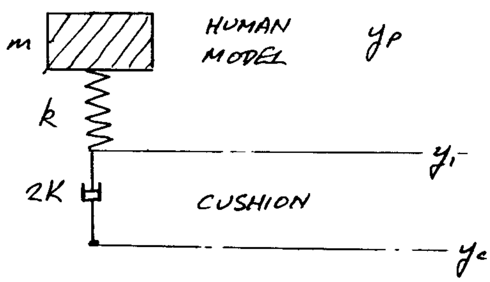


Figure 40. Wholly Viscous Restraint

The equation of motion for the spring-mass system is

$$\ddot{\delta} + \omega^2\delta = \ddot{y}_1 \qquad (57)$$

The upward force exerted by the damper is

$$F_D = 2K(\dot{y}_c - \dot{y}_1) = k\delta$$

i.e., $\dot{y}_1 = \dot{y}_c - k\delta/2K$

and $\ddot{y}_1 = \ddot{y}_c - \frac{k}{2K}\dot{\delta}$

Substituting for $\ddot{y}_1$ in equation 57.

$$\ddot{\delta} + \frac{k}{2K}\dot{\delta} + \omega^2\delta = \ddot{y}_c$$

or

$$\ddot{\delta} + \frac{\omega^2}{2c'} + \omega^2\delta = \ddot{y}_c \qquad (58)$$

where $c' = K/m$

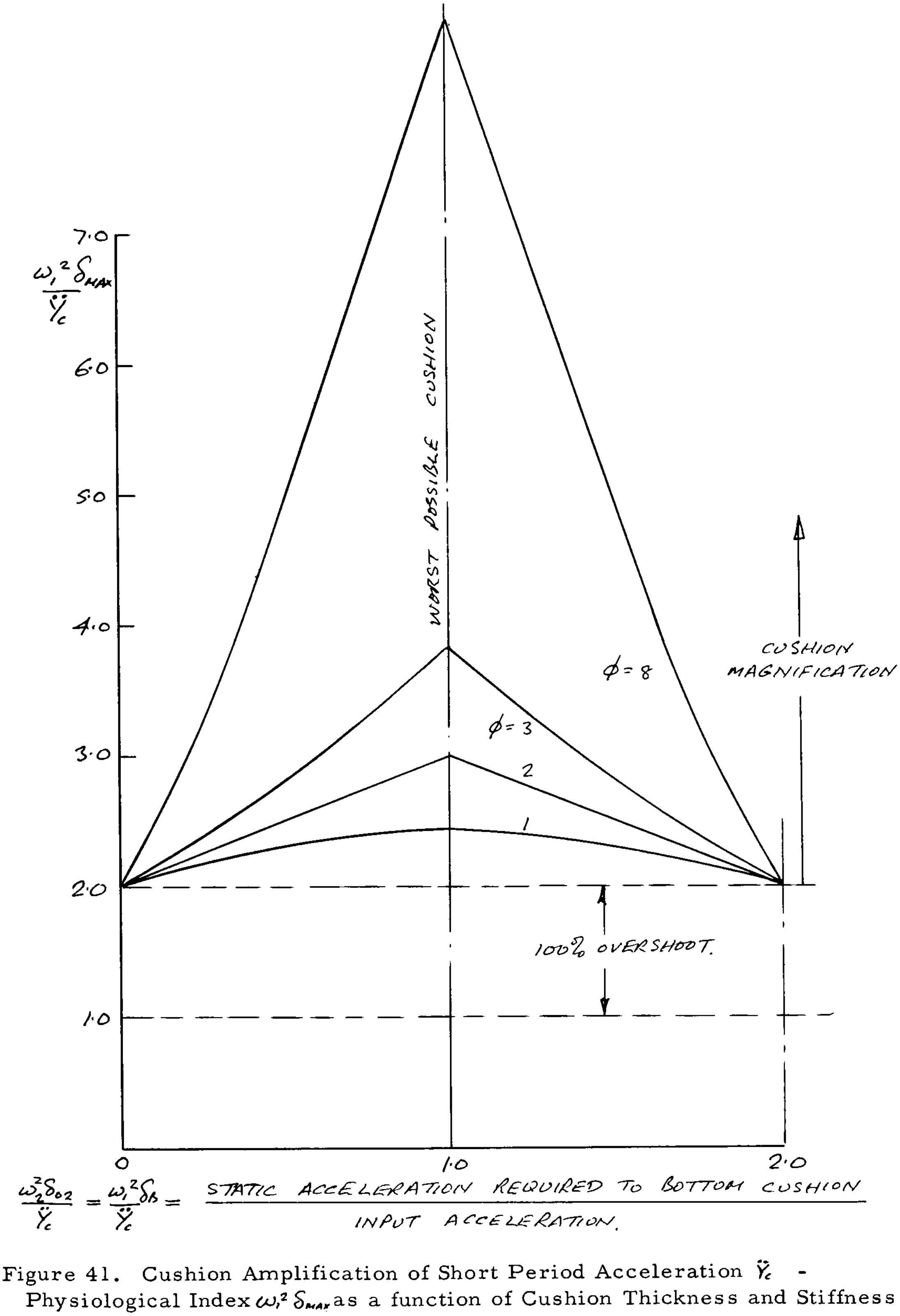


Figure 41. Cushion Amplification of Short Period Acceleration $\ddot{Y}_c$ - Physiological Index $\omega_1^2 \delta_{MAX}$ as a function of Cushion Thickness and Stiffness

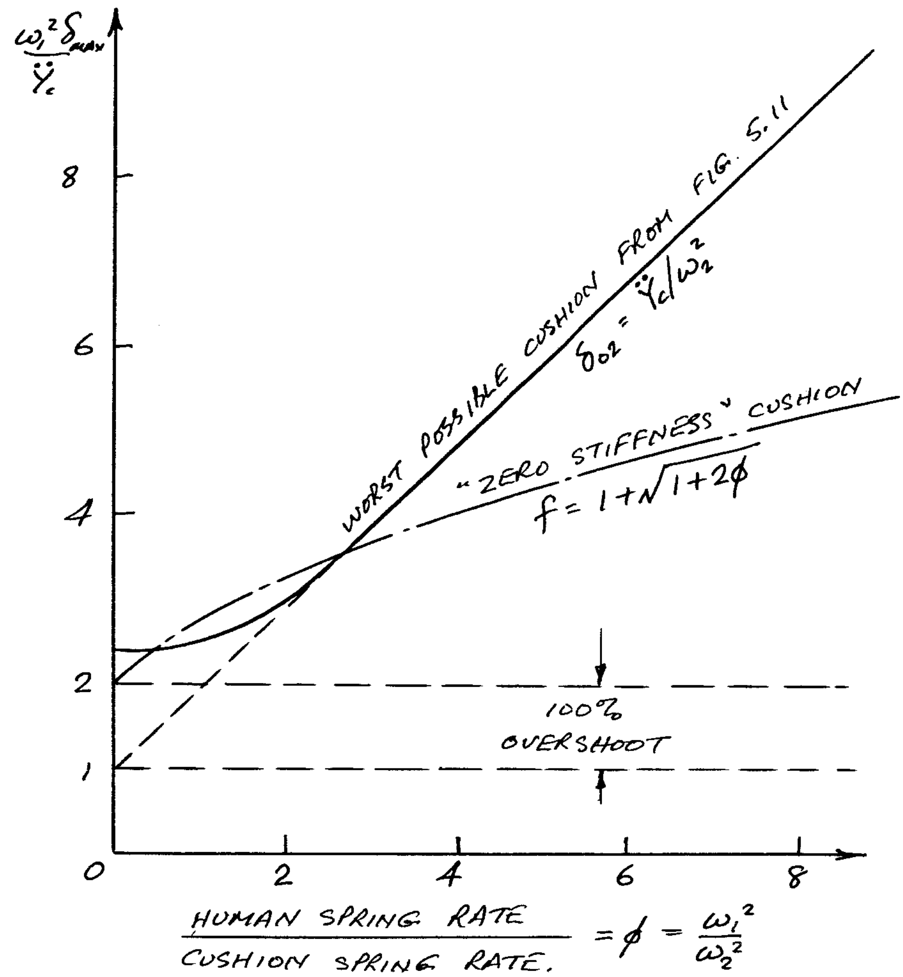


Figure 42. Variation of Acceleration Magnification with Cushion Stiffness for Worst Possible Cushion

This is a normal single degree of freedom equation, except that the normal damping coefficient

$$C = \frac{\omega^2}{2c'} \qquad \text{and} \qquad \bar{C} = \frac{C}{\omega} = \frac{\omega}{2c'} = \frac{1}{2\frac{c'}{\omega}}$$

Thus when $c'/\omega < \frac{1}{2}$ we have dead-beat motion, while for $c'/\omega > \frac{1}{2}$ the motion is sub-critically damped. Apart from these differences the solutions already developed can be directly applied in this case.

Linear Bottoming Cushion with Damping

This general case is illustrated in Figure 30, which constitutes a true two degree of freedom problem and in contrast to the idealizations previously considered, constitutes a sufficiently accurate model for use in analyzing experimental data, and for determining optimum restraint dynamics. Since theoretical investigations of this system have not been completed, and since it is undesirable to present many pages of mathematical analysis in a paper of this nature, no attempt will be made to discuss the dynamics of this system.

In general, however, the closed-form solutions are unwieldy, even for very simple inputs, and it is much more convenient to use an analog computer to study the effects of varying the various dynamic parameters of the restraint system.

Non-linear Restraint Systems

We have seen in Figure 35 that the non-linear behavior of a bottoming resiliency can be represented by a linear approximation, to a first order of accuracy. Many materials can be more accurately represented by a mathematical expression which is more complex than the linear equation

$$\text{force} = k\delta$$

Some of these expressions are given in Figures 43-45.

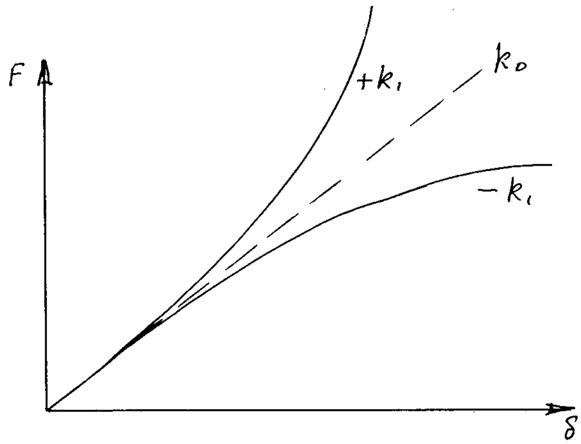


Figure 43. Cubic Elasticity

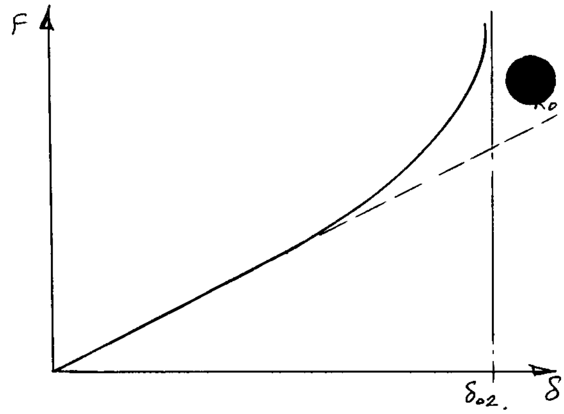


Figure 44. Tangent Elasticity

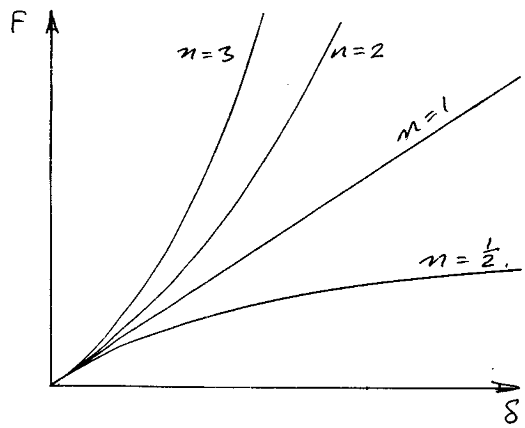


Figure 45. General Non-linear Spring of Degree

A cubic elasticity includes cushioning which does not bottom, but whose effective spring rate increases with deflection

$$\text{force} = k_o\delta + k_1\delta^3 \qquad (59)$$

Equations of motion containing this approximation can usually be solved by means of elliptic integrals. Note that it can also be used to represent a spring whose stiffness decreases with deflection by changing the sign of k_1.

A tangent elasticity represents very satisfactorily a bottoming cushion by the expression

$$\text{force} = \frac{2k_o\delta_{o2}}{\pi}\tan\left(\frac{\pi\delta}{2\delta_{o2}}\right) \qquad (60)$$

as shown in Figure 44. This expression is still simple enough for closed form solutions to be obtained without too much labor, and gives more accurate results than the linear approximation illustrated in Figure 35.

The general non-linear spring(8) can also be used to obtain many closed-form solutions, and is a useful approximation to many real cases. In this case the equation is

$$\text{force} = m \xi_n \delta^n \tag{61}$$

so that the equation of motion for a single degree of freedom system is

$$\ddot{\delta} + 2c\dot{\delta} + \xi_n \delta^n = f(t) \tag{62}$$

The writer(8) has discussed some early work in the solution of equation 62, while Mindlin(19) has considered simple impact solutions for cubic and tangent resiliences, although his results are not directly applicable to restraint problems because of the non-representative spring mass distributions used in his analysis. Mindlin(19) has also suggested a hyperbolic elasticity, but this has no applicability to restraint problems, except as a possible continuous function which represents the discontinuous behavior of crushable foam.

Non-linearities are also present in the damping coefficients of real materials, but the methods currently used for measuring energy dissipation are such that it is not easy to determine the precise nature of the non-linearity from the test data. Thus we are restricted to use of the linear values of β defined and presented on page 241.

The limitations involved in this approximation are not particularly serious at the present time, because of our lack of knowledge about material properties, and because of their wide variation with temperature. However, methods of extending present experimental research on material properties are suggested in the discussion of optimization of restraint (page 239).

Safety Nets

During the past few months a research program into the dynamics of personnel safety nets has been carried out at Frost Engineering, funded by the Rose Manufacturing Company of Denver. This program has already remitted substantial benefits in the design of new safety nets to tight procurement specifications, in the elimination of "cut and try" design techniques, and by pointing the way to more efficient materials and construction. It is of importance to the subject of this paper principally because a safety net can be regarded as a limit care of restraint, and because the non-linear theory has been developed beyond the fairly elementary levels so far considered.

It is obviously impossible to summarize the complete subject of this paper, and we must therefore confine ourselves to first principles and some important examples.

We assume the net illustrated in Figure 46 to be composed of b radial fibers (each with a cross section area a) so that the only loads carried are in a radial direction. The net is attached to a rigid frame via yielding elements which yield at a constant force R_y.

When a mass m impacts into the net with a velocity Δv the radial elements first deflect elastically until the load in them reaches the value R_y at which the yielding elements deflect. Further deflection is in the yielding elements, and this continues until all the initial kinetic energy $(\frac{1}{2} m \Delta v^2)$ of the mass has been absorbed.

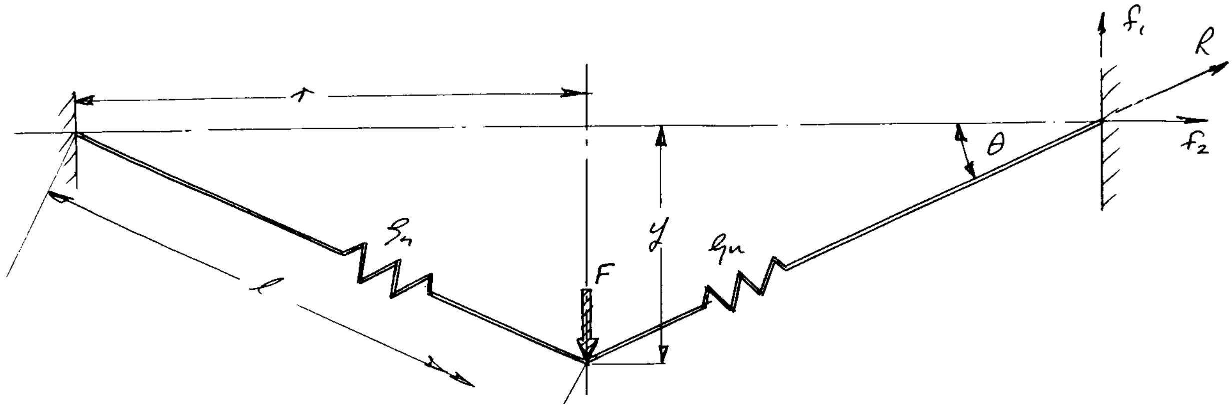


Figure 46. Net Geometry

The radial elements of the net are still stretched at this point, and now return to their original length, lifting the mass as they do so. The acceleration involved in this rebound cannot exceed the peak deceleration which occurs in arresting the mass, and the rebound height will depend upon the energy stored in the elastic phase of the net deflection, after taking hysteresis losses of the material into account. Net deflection causes a change in length e of the radial fibers, and the important parameters are suffixed as follows

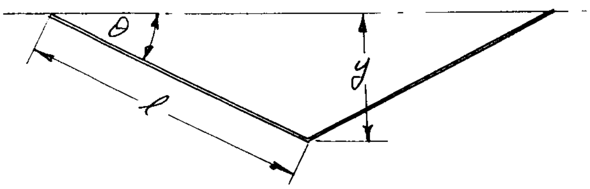


Figure 47

Initial values with zero load	$l_o\ y_o\ \theta_o$
Static value with mass lying stationary in the middle of the net	$l_s\ y_s\ \theta_s$
Values when the force R in the fibers is equal to the yielding value R_y	$l_y\ y_y\ \theta_y$
Maximum values reached in arresting the mass M	$l_m\ y_m\ \theta_m$

The energy absorbed in the elastic deflection of the net is

$$E_y = \int_{y_o}^{y_y} F\,dy$$

The energy dissipated during the yielding phase is

$$E_{m-y} = \int_{y_y}^{y_m} F\,dy$$

Thus the total energy balance equation is

$$\frac{1}{2} m \Delta v^2 = \int_{y_o}^{y_y} F\,dy + \int_{y_y}^{y_m} F\,dy.$$

By obtaining closed form solutions to the integrals we may determine y_m as a function of the other variables, and hence the peak deceleration G_{Max}.

General Theory

The force R in a radial element is

$$R_{y<y_r} = \xi_n \left(\frac{e}{l_o}\right)^n = \frac{\xi_n}{l_o^n}(l - l_o)^n$$
$$= \frac{\xi_n}{l_o^n}\left[\sqrt{y^2+r^2} - \sqrt{y_o^2+r^2}\right]^n \qquad (63)$$

The vertical force F associated with this net loading is

$$F = b f_1$$

But from Figure 46 $f_1 = R \sin\theta = Ry/\sqrt{y^2+r^2}$

$$\therefore\ F = bRy/\sqrt{y^2+r^2} \qquad (64)$$

$$= \frac{b\xi_n}{l_o^n}\frac{y}{\sqrt{y^2+r^2}}\left[\sqrt{y^2+r^2} - l_o\right]^n \qquad (65)$$

The energy absorbed is the integral of this equation with respect to y in accordance with the limits of equation 63. Unfortunately it is impossible to integrate in the general case of arbitrary n.

We can however integrate for specific values of n, and to do this most conveniently we use the non-dimensionalizing substitutions

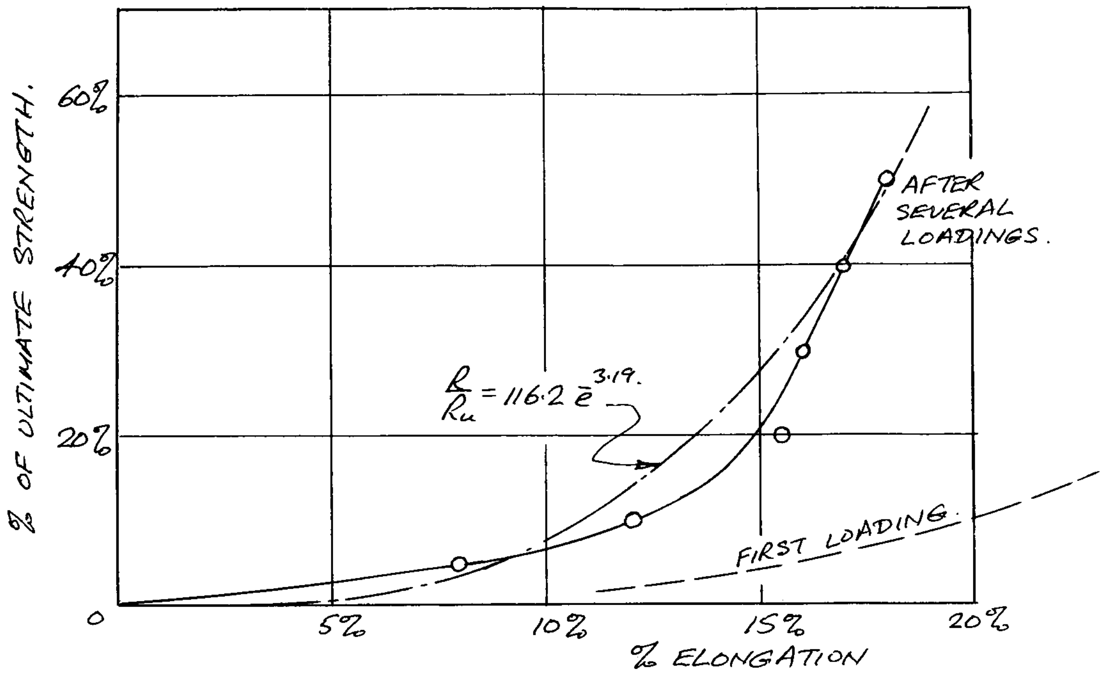
$$\bar{r} = r/l_o$$
$$\bar{y} = y/l_o$$
$$C_F = F/b\xi_n$$
$$C_{Ey} = E_Y/l_o b \xi_n$$

so that

$$C_F = \frac{\bar{y}}{\sqrt{\bar{y}^2+\bar{r}^2}}\left[\sqrt{\bar{y}^2+\bar{r}^2} - 1\right]^n \qquad (66)$$

$$C_{EY} = \int_{\bar{y}_o}^{\bar{y}_r} \frac{\bar{y}}{\sqrt{\bar{y}^2+\bar{r}^2}}\left[\sqrt{\bar{y}^2+\bar{r}^2} - 1\right]^n d\bar{y} \qquad (67)$$

For many net materials the exponent n is not yet adequately defined, one of the few examples where some test data exists being the nylon rope example given in Figure 48. The lack of this information does not inhibit us from utilizing the theory however, and the curves plotted in Figure 49, which are derived from equations 66 and 67, provides an excellent example of this. It is obvious that once the peak acceleration imposed by one net is known, we can use Figure 49 to predict the peak acceleration imposed by any other geometrically similar net, and for any other drop height. The error introduced by not knowing the correct value of the non-linear exponent n is evidently quite small and can be neglected for most practical purposes.



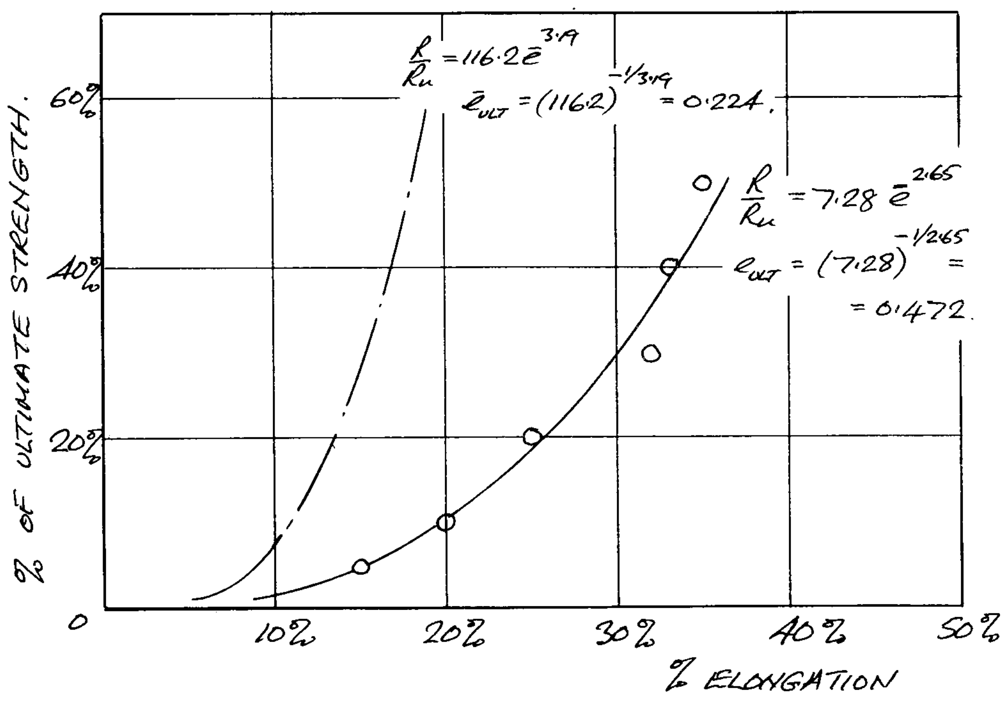


Figure 48. Load vs. Deflection for Nylon Rope

(Reference, Miskelly, Plymouth Cordage Co., Plymouth, Mass.)

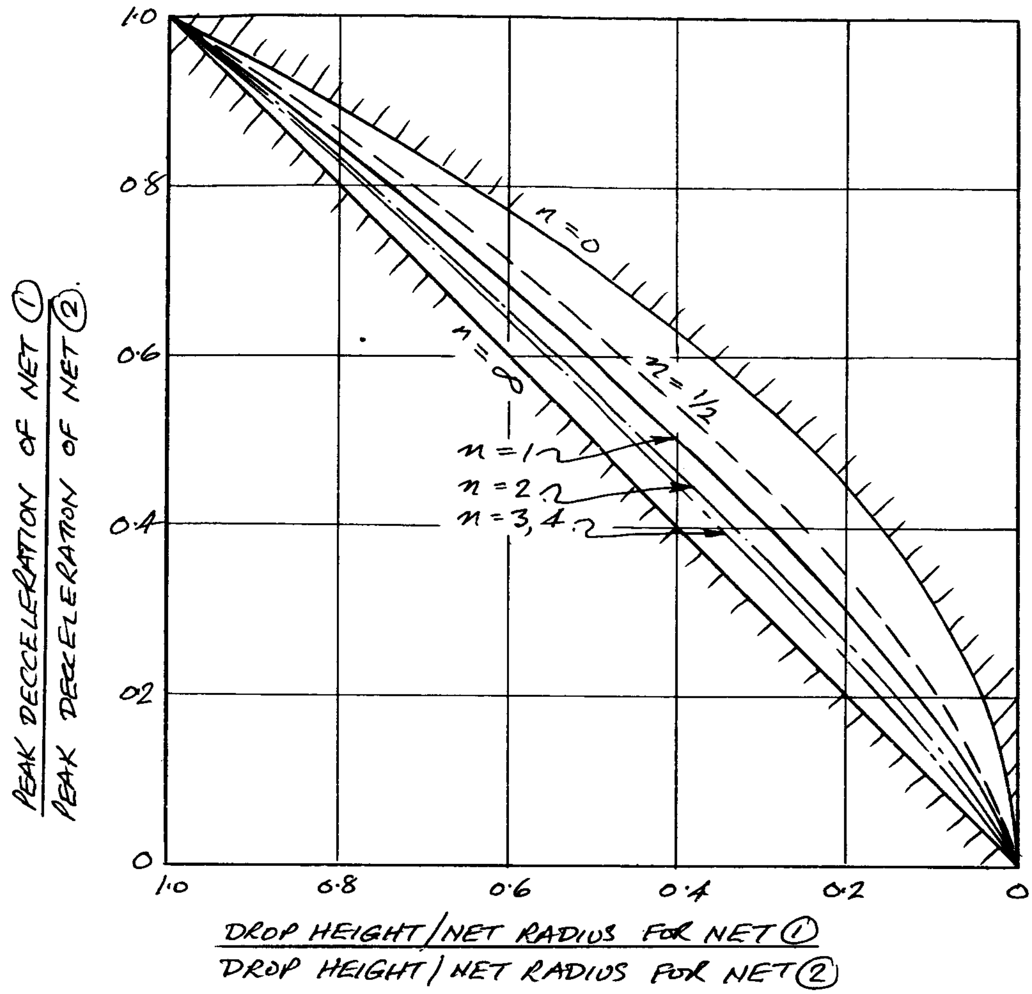


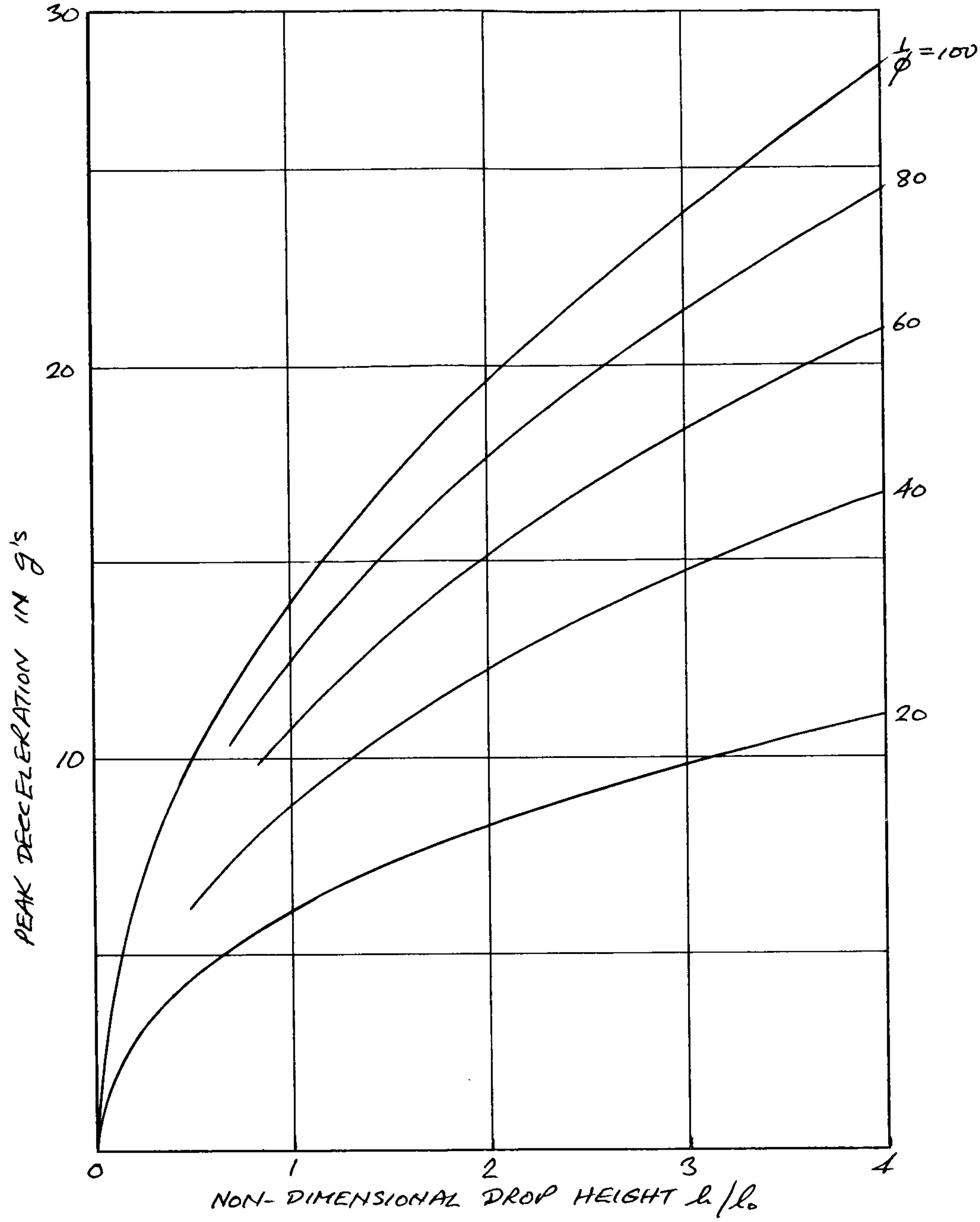
Figure 49. Variation of Peak Acceleration G_y in the Elastic Regime with Drop Height/Net Radius Ratio (h/l_o)

A net which yields at a constant force R_y in each member satisfies equation 64 as it stands, so that the peak deceleration which it imposes is

$$G_{MAX} = \frac{F_{MAX}}{gm} = \frac{\sqrt{\{1+\phi\bar{h}\}^2 - 1}}{\phi\{1+\phi\bar{h}\}} \tag{68}$$

where $\phi = gm/bR_y$

This equation is plotted in Figure 50.



$$\frac{1}{\phi} = \frac{\text{Yielding Force x Number of radial members}}{\text{Weight of Test Mass}}$$

$$\frac{h}{l_o} = \frac{\text{Drop Height}}{\text{Initial length of radial members}}$$

Figure 50. Exact Solution for Constant Force Yielding Net (Small Initial Sag)

Optimization of Restraint

We have seen that the dynamic characteristics of a restraint system can have a profound effect upon the acceleration tolerance of a vehicle's occupant, and that a restraint system which is beneficial for one type of acceleration input may prove to be lethal with another type. Thus almost any practical spring cushion, for example, will attenuate an impact acceleration and magnify a long duration acceleration.

A linear, bottoming, visco-elastic restraint device is dynamically much more complex than a simple spring cushion, so that the determination of optimum thickness, spring rate and damping coefficient for a particular type of acceleration input is correspondingly more difficult. The generation of closed-form solutions for impact and long duration accelerations (with variable rise time) will enable the optimum geometry to be determined for these simple cases, however, as well as for sinusoidal vibration. For more complex inputs we propose to use analog computer techniques as described in Reference 22.

It should be noted that many aerospace vehicles have a characteristic critical acceleration which is peculiar to the vehicle involved, and usually occur at some extreme points of the operational envelope. In such cases we are not concerned with maximum attenuation of the "normal" acceleration-time history, but in reducing the effects of the worst accelerations to the point where they are physiologically tolerable. Since these extreme accelerations may be very complex in shape, it is necessary to use either numerical or analog methods of solution. For optimization an analog is obviously far more economical than a digital computer, particularly in view of the inaccuracies inherent in our formulation of the dynamic coefficients of the restraint system.

Analog Computer Circuit

Since we do not yet know enough about human tolerance to acceleration to deal with arbitrary acceleration vectors, there is obviously no point in combining the lateral, transverse and spinal degrees of freedom. Thus we are concerned with a dynamic system of the type shown in Figure 51. The analog representation of this is relatively simple.

In the spinal case an additional degree of freedom is required to simulate the low frequency viscera mode shown on page 201.

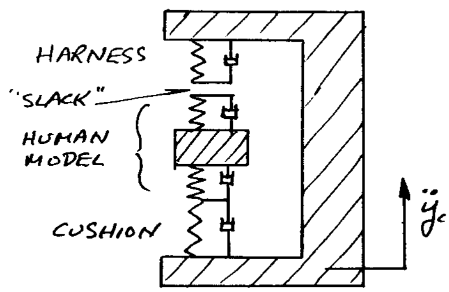


Figure 51. Dynamic System for Transverse Accelerations

No attempt will be made here to describe suitable analog techniques in detail, since this is the subject of Reference 22. It should be noted however that the cost of a general purpose computer which is capable of handling restraint dynamics is extremely high, and that, contrary to the impression sometimes conveyed by analog engineers, the development of a suitable and "bug-free" program must take several months. Thus it makes sound economic sense for a specialized "fixed-circuit" computer to be developed specifically for this problem: not only can it be expected to give relatively trouble-free operation in the hands of an engineer with no specialized analog experience, but since a number will be required, the capital outlay for each user will be only a small fraction of that required to program an in-house computer.

In addition, the "fixed-circuit" computer is instantly available, and can be used as desired, while it is not unusual to have to wait some weeks to gain access to a large general purpose machine.

Experimental Data

In the context of restraint the subject of experimental data falls naturally under three headings:

a. The fundamental dynamic properties of the materials available
b. Data relating to restraint system acceleration tests with anthropological dummies
c. Data relating to physiological experiments with human subjects

Although published summaries, notably that of Ungar and Hatch[13], enable us to summarize effectively the existing experimental data on available materials, the data relating to dummy and human subject acceleration tests will have to be gathered from the various agencies who are responsible for conducting these experiments.

The Dynamic Properties of Available Materials

The damping characteristics of existing materials are usually determined by applying a forced vibration to an appropriately supported test specimen, and measurements are taken to establish the relationship between mechanical impedance and frequency.

For convenience it is usual to define the modulus of elasticity in the complex plane:

i.e., $$E^* = E_1 + iE_2 = E(1 + i\beta) \qquad (69)$$

where E = the real part, corresponding to the modulus measured in static tests

$\beta = \dfrac{E_2}{E_1}$ = the "loss factor" associated with the modulus of elasticity

$i = \sqrt{-1}$

Similarly the complex shear stress may be expressed as

$$G^* = G_1(1 + i\beta) \qquad (70)$$

The relationship between the tensile and shear moduli is

$$G_1 = \frac{E_1}{2(1+\sigma)} \qquad (71)$$

where σ = Poisson's ratio

$\simeq$ 0.5 for high damping materials

As a first approximation, the values of the loss coefficient β can be taken to be the same for both shear and tension. Typical values for both E_1, and β are given in Table 1.

TABLE 1

Dynamic Coefficient Ranges for Common Materials

	β	E x 10^6 (lb/in^2)
Metals	.0001 - .001	5.0 - 35.0
Glass	.001 - .005	8.0 - 12.0
Concrete, brick	.001 - .01	1.0 - 10.0
Sand, granular media	.01 - .05	0.1 - 3.0
Wood, cork	.01 - .02	0.5 - 2.5
Rubber, plastics	.001 - 10.0	0 - 0.5

It is obvious that the most attractive materials, from the point of view of energy dissipation, are rubber and plastics. In practice these can be tailored to give almost any desired combination of stiffness and damping, although these parameters are also functions of temperature and loading history.

In Tables 2 - 4 (after Ungar and Hatch[13]) some existing data is summarized from experiments in which mechanical impedance was determined by the imposition of a forced sinusoidal vibration.

TABLE 2

Materials for which Extensive Data are Available

Material	Max. Loss Factor β Max.	Position of Freq. CPS	β Max. Temp. ^oC	Shear Modulus G, At β Max. PSI
Plasticized polyvinyl acetate	2.6	50	5	2,180
Polystyrene	2.0	2000	140	14,500
Polyisobutylene	2.0	6000	25	290
Polysulfide rubber (Thiokol RD)	1.9	7	5	10,000
Polyvinyl chloride	1.8	20	92	3,330
Buna N (Type B-1), vulcanized	1.5	4000	20	1,450
Polymethyl methacrylate	1.5	1200	142	14,500
Plasticized PVC (Koroseal)	1.45	660	50	3,470
Polyester	1.1	200	108	10,100
Polytetrafluoroethylene (Teflon-TFE)	1.0	400	23	5,800
Hard rubber	1.0	40	60	10,100
Nit[illegible] rubber	0.8	1800	20	15,900
Urethane rubber (Shore 80A)	0.8	30	-8	2,320
Filled rubber	0.5	2500	22	33,300
Urethane rubber (Shore 94A)	0.4	140	-23	43,500
Aquaplas (water soluble)	0.38	1000	40	24,200

TABLE 3

Materials for which Data are Available Only at Low Frequencies

Material	Max. Loss Factor β Max.	Temp. Range C° (for β Max)	Test Freq. CPS	Shear Modulus G, At β Max. PSI
Polyvinylcarbazol	1.6	200/220	10.0	1,200
Polyvinyl-N-butyl ether	1.6	-40/-20	0.8	1,500
Polyvinyl-isobutyl ether	1.6	-15/10	1.2	1,500
Polyvinyl-tert-butyl ether	1.6	70/90	1.7	1,500
Butyl rubber (unvulcanized)	1.2	-20/20	1.0	15,000
Neoprene GN-50 (vulcanized)	0.5	-20/5	150.0	5,800
Natural rubber (high mol wt)	0.5	-70/-50	10.0	1,500
GR-S rubber	0.5	-60/-40	10.0	7,[illegible]0
Polytrifluorochloroethylene	0.42	90/120	3.3	11,600
Polyvinylfluoride	0.36	20/60	1.7	5,800
Polyethylene	0.23	30/80	12.0	2,900
Buna N (B-5), carbon-filled, vulcanized	0.15	0/10	50.0	29,000

TABLE 4

Materials for which Data are Available Only at Room Temperature

Material	Max Loss Factor β Max CPS	Frequency for β Max CPS
1. Polysulfide rubber (Thiokol H-5)	5.00	1,000
2. Butyl rubber (Enjay 9-262-4)	4.02	3,100
3. Urethane rubber (Disogrin IDSA 9250)	2.59	3,000
4. Butyl rubber (Enjay 9-262-1)	2.56	3,000
5. Butyl rubber (Enjay 9-262-3)	2.20	3,000
6. Polyvinyl butyral	2.0	2
7. 3M tape adhesive No. 466	1.82	1,000
8. Butyl gum (Type U-50), vulcanized	1.80	10,000
9. Buna N (Type B-5), carbon-filled, vulcanized	1.60	10,000
10. GR-S rubber	1.60	10,000
11. Urethane rubber (82% Pb powder filler)	1.40	3,000
12. Fluoro rubber (3M No. 1F4)	1.30	4,000
13. Neoprene GRT	1.18	1,775
14. Polyvinyl chloride acetate	1.14	100
15. Neoprene (Type GN-50), vulcanized	1.10	10,000
16. Buna N (Type B-0), unvulcanized	1.00	1,000
17. Hycar 1014	0.76	1,850
18. 3M camping tape adhesive, No. 435	0.73	4,000
19. Neoprene (Type CG-1), vulcanized	0.60	10,000
20. Fluro silicone rubber (Silastic LS53)	0.56	4,000
21. Polysulfide rubber (Thiokol ST)	0.53	1,300
22. Urethane rubber (Disogrin IDSA 7560)	0.51	4,250
23. GR-S rubber (23.5% styrene)	0.47	1,700
24. GR-S rubber (3.0% styrene)	0.38	1,400
25. GR-S (Type S-50) carbon-filled, vulcanized	0.35	10,000
26. Silicone gum (Linde No. Y-1032)	0.33	400
27. Natural gum rubber (Type N-1)	0.30	1,000
28. GR-S rubber (Krylene)	0.30	10,000
29. Hevea rubber (filled)	0.25	10,000
30. Hevea rubber (vulcanized)	0.20	10,000
31. Natural rubber (Tire tread stock)	0.20	500

TABLE 4

Materials for which Data are Available Only at Room Temperature (Continued)

Number	Loss Factor at Five Frequencies (C.P.S.) 1	10	100	1,000	10,000	Test Temp. C°	Shear Modulus G, At β Max PSI
1.	0.17	0.50	1.20	5.00		25 ± 50	1,500
2.			0.40	0.92	2.0	21	1,500
3.			0.10	0.13	0.50	25	2,900
4.			0.55	0.84	1.80	21	2,900
5.			0.45	0.68	1.40	24	2,900
6.		0.14	0.06	0.06		30 ± 70	29,000
7.		1.00	1.17	1.82	0.80	25	[illegible]
8.				1.20	1.80	20	1,000
9.		0.40	0.75	1.20	1.60	20	1,500-4,400
10.		0.20	0.50	1.00	1.60	20	1,500
11.			0.40	0.49		25	2,500
12.		0.30	0.46	0.79		45 ± 55	17
13.			0.60	1.01		20	220
14.			1.14	0.73	0.90	40 ± 20	580
15.				0.70	1.10	20	290
16.		0.50	0.65	1.00		15	1,200
17.			0.40	0.63		20	200
18.		0.50	0.58	0.67	0.70	10 ± 50	37
19.				0.02	0.60	20	4,350
20.		0.10	0.16	0.46		30 ± 70	87
21.			0.40	0.51		20	160
22.			0.10	0.22		25	1,500
23.			0.30	0.39		20	200
24.			0.25	0.37		20	13
25.				0.20	0.35	20	435
26.		0.15	0.20	0.40		60 ± 40	2.5
27.	0.04	0.08	0.14	0.30		20	150
28.	0.11	0.12	0.13	0.15	0.3	20	100
29.	0.11	0.11	0.11	0.14	0.25	20	870
30.	0.03	0.03	0.03	0.05	0.20	20	870
31.	0.16	0.18	0.20			25 ± 50	150

Acceleration Tests with Anthropological Dummies

Many agencies have conducted acceleration tests in which an anthropological dummy has been accelerated via a seat and restraint system. If we replace the human dynamic model in Figure 51 by a single rigid mass, and if the dummy is "stiff" in relation to the dynamic input, then the equations developed will enable its acceleration to be predicted. More importantly, if the dynamic characteristics of the restraint

system are unknown, then the analog computer circuits which have been developed can be used in reverse to determine the effective damping and spring rate of both the cushion and harness.

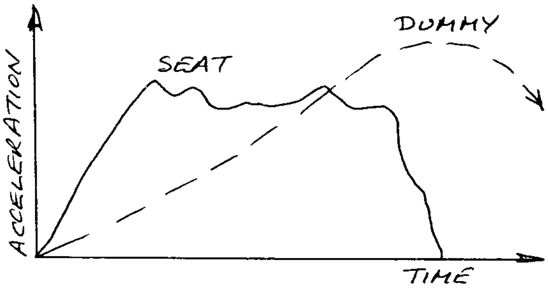


Figure 52. Typical Dummy Test Data

Particularly in the case of harness dynamics therefore, the analysis of existing experimental data will be of considerable value in determining the gross dynamic characteristics of a restraint system, although the overall accuracy of this type of analysis is not expected to be high.

Acceleration Tests with Human Subjects

Most acceleration experiments with human subjects involve some degree of restraint resiliency, and it is possible that in some the true physiological effects of the known acceleration input are completely masked by the restraint system dynamics. If the gross dynamic characteristics of the restraint are known for a particular experiment, then it is obvious from the foregoing sections that the acceleration-time history actually imposed by the restraint system can be calculated.

Future Experimental Research

While it is difficult to predict what experimental research will be required until a comprehensive theoretical study program has been completed, it is evident that a real need exists to study the behavior of visco-elastic materials in more detail than is currently achieved with mechanical impedance measurements. Ideally we should be able to determine the variation of force with both extension δ and extension rate $\dot{\delta}$ for various temperatures and degree of pre-loading. To accomplish this we need some form of linear test device, possibly one of the devices suggested diagrammatically in Figure 54. Both these devices can be instrumented to give a continuous load-time history, but device (a), equipped with a waxed scale to show maximum deflection of the test mass, can also be used to determine gross dynamic characteristics of a material without the need for any additional instrumentation. In other words, if we assume that the test mass and specimen of Figure 54 (a) correspond to a single-degree-of-freedom system with linear damping, and if we also know the static properties of the test specimen, then the maximum deflection of the mass for a known input acceleration enables us to calculate the equivalent linear damping coefficient.

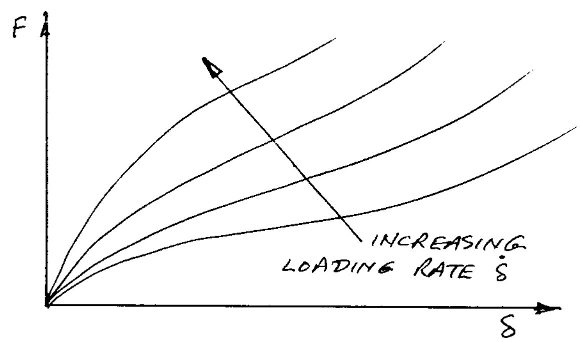


Figure 53

The impact tester in Figure 55 is even simpler to operate, since we know the impact energy content from the weight and drop height of the hammer. By using a waxed scale or a piece of modelling clay to determine maximum deflection of the test specimen we can plot energy absorption against deflection, and determine the maximum

energy absorption which is achieved when the test specimen effectively bottoms out. The impact velocity for a given impact energy can be varied by varying the weight of the hammer head.

In general terms, it is anticipated also that a number of anthropological dummy tests will have to be made at some suitably equipped facility in order to determine the dynamic characteristics of some typical restraint harnesses, and to relate this to the measured properties of the material(s) from which the harnesses are manufactured.

Finally, of course, it would be highly desirable to accelerate some human test subjects in restraint systems which have been designed in accordance with the principles presented in this report, under conditions which permit a comparison to be made with existing harnesses.

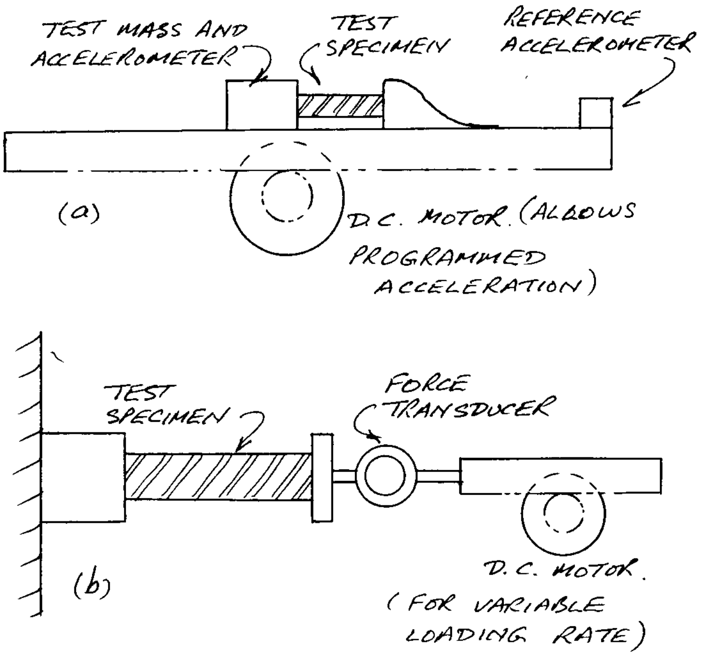


Figure 54. Two Methods of Measuring the Dynamic Properties of Visco-elastic Materials

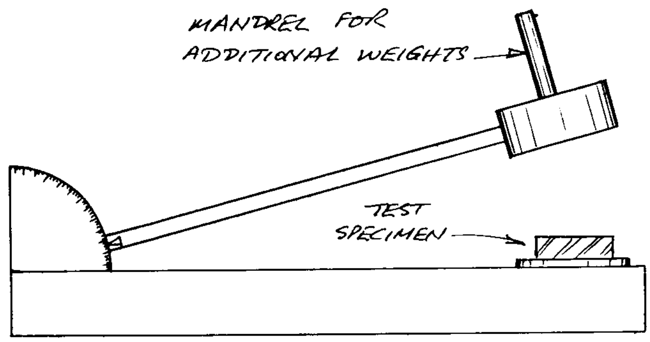


Figure 55. Simple Device for Measuring Material Properties under Impact

Proposed Research Programs

Proposals Already Made to NASA by Frost Engineering

Study of Restraint Dynamics

A technical proposal was submitted to NASA in August, 1961, the financial proposal following in September. The following summary of the program objectives is quoted from the financial proposal.

"To develop the theory of restraint system dynamics, using mainly linear theory, and to present the results in a report which will have the following main divisions:

1. A simple physical description of the dynamics of human tolerance to short period acceleration.

2. A simple physical description of the influence of restraint system dynamic characteristics upon the tolerability of given input accelerations.

3. The development of linear restraint system theory, and the determination of equivalent linear equations and non-linear systems.

4. The use of theory to define optimum dynamic characteristics for restraint systems.

5. The development of suitable analog circuits, and their use on the Frost analog computer to produce plots of optimum values for the individual variables, such as cushion stiffness, depth, and hysteresis damping, and the implications of these optimums in terms of hardware requirements.

6. A summary of all the available information on visco-elastic materials suitable for use in restraint systems.

7. The extension of existing dynamic models to include "long period" acceleration tolerance of the order of minutes in duration.

8. A digital computer program which embodies the theory developed.

9. Methods of estimating the influence of restraint in physiological experiments, so that the data obtained can be reduced to the common level of the acceleration actually experienced by the test subject's body, rather than the input to the restraint system, as is presently the case.

10. A review of the immediate requirements for further experimental work, and a proposed long term research program."

It will be noted that item 5 is concerned with "..... the development of suitable analog (computer) circuits." Reference 22 shows that this has already been accomplished, at least in part, so that this item is somewhat out-dated. The reason for this apparent inconsistency is that the decision to design and build a computer on a company-funded basis was not taken until September 9, when the invitation to attend this symposium was received.

Acceleration Handbook Proposal

In most technical disciplines it is possible to refer to an authoritative "handbook" which contains the generally agreed answers to problems already solved and approved definitions and techniques. For example, no aircraft stressman could function effectively without his copy of ANC-5 ("Strength of Metal Aircraft Elements") and no propeller designer would willingly be without ANC-9. Not only do such handbooks save a great deal of work on the part of the engineers using them, but they also define a common language and scale of values, so that the customer does not have to be told how the answers were achieved, and upon what assumptions they rest.

When such a facility is absent the expenditure of man-hours in semantic and communication problems alone are enormous; as can be seen by watching the process whereby the Air Force and a contractor try to decide whether the accelerations imposed on an aerospace vehicle occupant are tolerable. Thus a very good case exists for preparing a "Handbook of Human Tolerance to Acceleration," and obtaining general approval of its contents by all agencies, contractors and research teams working in this field.

Such a handbook would have to be kept up to date on an annual basis, at least initially, and would be written at three levels.

1. An elementary level.

2. An intermediate level in which the physical picture is discussed in some detail.

3. A mathematical level, in which rigorous proofs are advanced for the techniques proposed in the earlier sections.

Although the core of this handbook would be the latest information on the gross limits of human tolerance, and the models which describe this information mathematically, no attempt would be made to include a summary of physiological research, except in the general manner used by Goldman and von Gierke[7] in order to give essential background.

Standard numerical, digital and analog programs for determining "Physiological Index" would of course be essential elements of such a handbook, in order that each contractor could automatically assess accelerations imposed by his designs and quote merely the maximum P.I. figure to the procuring agency.

Future Research Work

It is evident from the section titled "The Dynamics of Head Restraint," of this paper, that a great deal of theoretical work remains to be done in connection with head acceleration tolerance levels, and when sufficient time becomes available to study more fully the work already carried out we hope to make a proposal in this area.

Apart from this we could list perhaps fifty speculative areas in which research might yield profitable results, but it is felt that little benefit would be derived from such a listing. Despite the fact that most research is programmed, and despite the recent developments in management control such as PERT, we still believe that true research cannot be programmed. To quote from Professor Parkinson's most rec[ent] study[23],

> "No king or minister could have instructed Newton to discover the law of gravity, for they did not know and could not have known that there was any such law to discover. No treasury official told Fleming to discover penicillin. Nor was Rutherford instructed to split the atom by a certain date, for no politician of his day and scarcely any other scientist would have known what such an achievement might imply, or what purpose it would serve. Discoveries are not made like that.

They are the result, as often as not, of someone wandering off his own line of research, attracted by some phenomenon hitherto unnoticed or suddenly seen in a new light."

A second aspect of applied research, which does not seem to have received enough attention is the importance of exposing the research worker to practical problems. From a purely personal point of view the concepts outlined in this paper would have never occurred to the writer if he had not been connected with work on the Stanley B-58 escape capsule, and if Galen Holcomb had not regularly exposed him to the physiological problems connected with it. Thus to be truly effective, the applied research worker should be constantly exposed to practical problems in his field, and yet have complete intellectual freedom at the same time. At any given time he is usually quite unable to predict where he will be in six months time, and may be completely wrong if he attempts to do so. If the writer's work had been tied down contractually to the lines of investigation of a year ago, for example, most of the effort would have been wasted.

With these reservations, however, there are four general areas which do not appear to be receiving sufficient attention at the present time, in addition to the head tolerance work mentioned at the beginning of this section.

Impact Testing with Human Subjects

The tolerance limits defined in Figures 11 and 12 indicate that the impact region is for durations

$$\Delta t < .007 \text{ secs (Spinal)}$$
$$\Delta t < .03 \text{ secs (Transverse)}$$

Some transverse experiments have been carried out on accelerators with acceleration durations of less than these critical durations, but such a procedure is very costly. The same data can be obtained by simply dropping the test subject, properly supported in a couch or chair, onto a rigid floor surface, and a large number of experimental points can be obtained in this way; particularly since no instrumentation is needed. The work of Kornhauser with mice for example, is wholly admirable, and should give us valuable insight into the statistical variations to be expected. An important test point was also obtained this way by Dr. Swearingen, and was used in the construction of Figure 11.

This type of work could best be carried out by one of the military research agencies, the object being to first determine the critical drop height (or velocity change) of as many as 100 subjects, the term "critical" referring to some subjective end-points which only the medical workers in this field can safely define.

Integrated Research

From a long term point of view there can be little doubt that the "integrated research" effort achieved when a team of medical, engineering and mathematical workers are employed on a continuing program, is by far the most effective means of increasing our knowledge. Perhaps the best example of this is the Bio-Medical Laboratory at ASD, where Drs. von Gierke, Coermann, and others of their team are doing such excellent work. Too many of our research programs today are being undertaken by workers with a capability in only one of the several disciplines involved, with the result that their work loses much of its value. To take only one example, the task of

instrumenting a physiological experiment is one which taxes the ingenuity of a capable electronics engineer of wide experience and considerable mathematical insight. How then can we believe the results of experiments which report instrumentation simply in the words "....accelerometers were mounted...."?

"Odd-ball" Research

As a comparative newcomer to the field of acceleration tolerance the writer first heard of the work of Kornhauser(9, 10) indirectly, and formed the impression that most workers regarded his theories as extremely peculiar. Yet today many of us would be inclined to rate his work very highly indeed, and to deplore the fact that he is not able to proceed at an accelerated pace.

It therefore seems important that we should find a way of funding programs which are not generally approved of by the majority of workers in the field; both the experience workers who are interested primarily in their own chosen fields, and those who Professo Parkinson(23) has so aptly christened the "Abominable No-Men."

Statistical Techniques in Data Evaluation

Our main concern in acceleration tolerance work is with healthy young males who have passed one or more rigorous medical examinations, so that we can expect the variability of tolerance levels to be less than for a broad sample of the population. Nevertheless a variation will exist, so that our single-valued tolerance envelopes should actually be replaced by a number of probability envelopes, defined by the percentage of the population to which each limit refers. It is obvious that, in the impact range at least, the mice experiments of Kornhauser(10) will give us valuable insight into the standard deviation to be expected, although the absolute scatter will naturally be larger than for carefully selected human subjects.

These considerations alone would be sufficient reason for introducing the use of statistical and probability data analysis techniques, but an even more cogent reason is the possibility of utilizing the great mass of existing acceleration research data in which no definable end-point was reached. This possibility is outlined in some detail in Appendix B of Reference 22, where it is shown that the theory of "successful successive tests" may help to define end-points, even though none of the tests approached an end-point.

Apart from increasing the precision of our tolerance limit estimates from existing data, it is felt that these new techniques will simplify the problem of planning new experiments and may possibly permit "non-destructive testing" to be used almost exclusively for human subjects.

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APPENDIX A.

Effect of Linear Rise Time Upon Response of a Simple Spring-Mass System

The equation of motion written in terms of spring deflection δ, is

$$\ddot{\delta} + \omega^2 \delta = \ddot{y}_c \qquad \text{(A.1)}$$

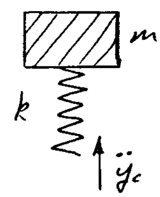

Figure A.1

In Laplacian notation the solution for δ is therefore

$$\bar{y}(p) = \frac{\mathcal{L}(\ddot{y}_c)}{p^2+\omega^2} + \frac{p y_0}{p^2+\omega^2} + \frac{y_1}{p^2+\omega^2} \qquad \text{(A.2)}$$

so that $\delta = (\delta)_0 \cos\omega t + \frac{(\dot{\delta})_0}{\omega}\sin\omega t + \mathcal{L}^{-1}\left[\frac{\mathcal{L}(\ddot{y}_c)}{p^2+\omega^2}\right]$ (A.3)

where $(\delta)_0 = \delta$ at time $t = 0$

$(\dot{\delta})_0 = \dot{\delta}$ at time $t = 0$

Thus if we know the Laplace transform of the input pulse, we can rapidly determine the system's response to it. For the acceleration-time history depicted in Figure A.2

$$\ddot{y}_c = \ddot{Y}_c \frac{t}{t_R} \qquad (t < t_R)$$

$$= \ddot{Y}_c \qquad (t > t_R)$$

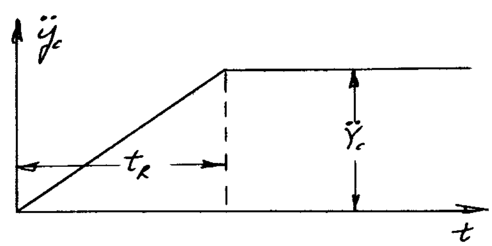

Figure A.2

The transformation is

$$\mathcal{L}(\ddot{y}_c) = \frac{\ddot{Y}_c}{t_R}\frac{1}{p^2}\left(1 - e^{-pt_R}\right) \qquad \text{(A.4)}$$

$$\therefore \frac{\mathcal{L}(\ddot{y}_c)}{p^2+\omega^2} = \frac{\ddot{Y}_c}{t_R}\left\{\frac{1}{\omega^2 p^2} - \frac{1}{\omega^2(p^2+\omega^2)}\right]\left(1 - e^{-pt_R}\right)$$

$$\mathcal{L}^{-1} = \frac{\ddot{Y}_c}{\omega^3 t_R}\left\{\omega t - \sin\omega t\right]\left[1 - H(t - t_R)\right] \qquad \text{(A.5)}$$

where H is the Heaviside operator

$H(t-t_R) = 0$ for $t < t_R$ and 1.0 for $t > t_R$

Thus the complete solution of equation A.3 is

$$\delta = (\delta)_0 \cos \omega t + \frac{(\dot{\delta})_0}{\omega} \sin \omega t + \frac{\ddot{Y}_c}{\omega^3 t_R}\{\omega t - \sin \omega t\}\{1 - H(t - t_R)\} \qquad \text{(A.6)}$$

for $(\delta)_0 = (\dot{\delta})_0 = 0$ we can plot

$$\frac{\ddot{y}_P}{\ddot{Y}_c} = \frac{\omega^2 \delta}{\ddot{Y}_c} = \frac{1}{\omega t_R}\{\omega t - \sin \omega t\}\{1 - H(t - t_R)\} \qquad \text{(A.7)}$$

For $t > t_R$

$$\left(\frac{\ddot{y}_P}{\ddot{Y}_c}\right)_{t > t_R} = 1 - \frac{\sqrt{2}}{\omega t_R}\sqrt{1 - \cos \omega t_R}\, \sin(\omega t + \phi) \qquad \text{(A.8)}$$

where $\sin \phi = \sin \omega t_R / \sqrt{2(1 - \cos \omega t_R)}$

By inspection the peak value is

$$\left(\frac{\ddot{y}_P}{\ddot{Y}_c}\right)_{\substack{MAX \\ t > t_R}} = 1 + \frac{\sqrt{2}}{\omega t_R}\sqrt{1 - \cos \omega t_R} \qquad \text{(A.9)}$$

For small values of ωt_R, the right-hand side of equation A.9 reduces to 2.0, which is the classic result for zero rise time. As Figure A.3 indicates, a rise time which is less than $1/\omega$ may be regarded as zero, for all practical purposes, while we may generally assume that for $t_R > 10/\omega$ there is no dynamic overshoot.

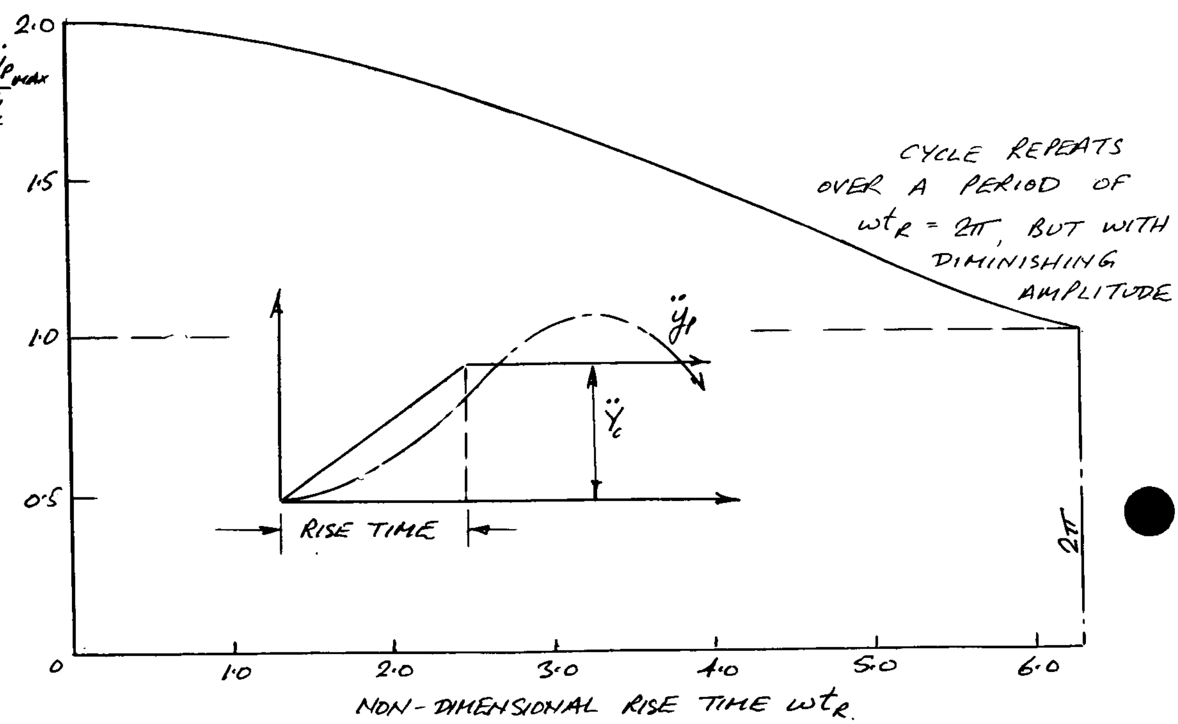

Figure A.3. Effect of Rise Time t_R on Response of Simple System to Acceleration

APPENDIX B.

Basic Equations For a Two Degree of Freedom Model

The equations of motion for the linear, undamped system in Figure B.1 are

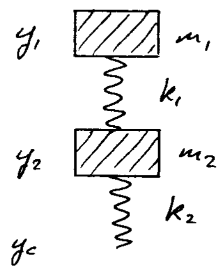

Figure B.1

$$\left.\begin{aligned} m_1\ddot{y}_1 &= -gm_1 + k_1\delta_1 \\ m_2\ddot{y}_2 &= -gm_2 + k_2\delta_2 - k_1\delta_1 \end{aligned}\right\} \qquad \text{(B.1)}$$

$$\left.\begin{aligned} \ddot{y}_1 - \omega_1^2\delta_1 &= -g \\ \ddot{y}_2 - \omega_2^2\delta_2 &= -g - \varsigma\omega_1^2\delta_1 \end{aligned}\right\} \qquad \text{(B.2)}$$

where $\varsigma = m_1/m_2$

now $\delta_1 = \delta_0 - (y_1 - y_2) \quad \therefore \quad \ddot{y}_1 = \ddot{y}_2 - \ddot{\delta}_1$

also $\ddot{y}_2 = \ddot{y}_c - \ddot{\delta}_2 \quad \therefore \quad \ddot{y}_1 = \ddot{y}_c - \ddot{\delta}_2 - \ddot{\delta}_1$

Substituting in equation B.2

$$\begin{aligned} \ddot{\delta}_1 + \omega_1^2\delta_1 + \ddot{\delta}_2 &= (\ddot{y}_c + g) \\ \ddot{\delta}_2 + \omega_2^2\delta_2 - \varsigma\omega_1^2\delta_1 &= (\ddot{y}_c + g) \end{aligned} \qquad \text{(B.3)}$$

Obviously we may write $\ddot{y}_c$ for $(\ddot{y}_c + g)$ so long as we remember to include g in our definition of $\ddot{y}_c$. Also, by equating equations B.3, we can express δ_1 as a function only of δ_2

i.e., $\ddot{\delta}_1 + \omega_1^2(1+\varsigma)\delta_1 = \omega_2^2\delta_2$ (B.4)

Laplace Transform of Equations

The equations

$$\begin{aligned} \ddot{\delta}_1 + \omega_1^2\delta_1 + \ddot{\delta}_2 &= \ddot{y}_c \\ \ddot{\delta}_2 + \omega_2^2\delta_2 - \varsigma\omega_1^2\delta_1 &= \ddot{y}_c \end{aligned} \qquad \text{(B.5)}$$

become, in Laplacian form, for zero initial conditions

$$\begin{aligned} p^2\bar{y}(p)_1 + \omega_1^2\bar{y}(p)_1 + p^2\bar{y}(p)_2 &= \mathcal{L}(\ddot{y}_c) \\ p^2\bar{y}(p)_2 + \omega_2^2\bar{y}(p)_2 - \varsigma\omega_1^2\bar{y}(p)_1 &= \mathcal{L}(\ddot{y}_c) \end{aligned} \qquad \text{(B.6)}$$

Dividing by the $\bar{y}(p)_1$, coefficients

$$\begin{aligned} y(p)_1 + \frac{p^2}{p^2+\omega_1^2}y(p)_2 &= \frac{\mathcal{L}(\ddot{y}_c)}{p^2+\omega_1^2} \\ -y(p)_1 + \frac{1}{\varsigma\omega^2}(p^2+\omega_2^2)y(p)_2 &= \frac{\mathcal{L}(\ddot{y}_c)}{\varsigma\omega_1^2} \end{aligned}$$

Adding, we have

$$y(p)_2\left\{\frac{p^2}{p^2+\omega_1^2}+\frac{1}{\xi\omega_1^2}(p^2+\omega_2^2)\right\}=\mathcal{L}(\ddot{y}_c)\left\{\frac{1}{p^2+\omega_1^2}+\frac{1}{\xi\omega_1^2}\right\}$$

Multiply both sides by $\xi\omega_1^2(p^2+\omega_1^2)$

$$\bar{y}(p)_2=\mathcal{L}(\ddot{y}_c)\left[\frac{\xi\omega_1^2}{(p^2+\omega_1^2)(p^2+\Omega^2)}+\frac{1}{p^2+\Omega^2}\right] \tag{B.7}$$

where $\Omega^2=\omega_2^2+\xi\omega_1^2(1-\omega_1^2)$

$$=\omega_2^2\left\{1+\xi\frac{\omega_1^2}{\omega_2^2}(1-\omega_1^2)\right\} \tag{B.7a}$$

Resolving equation B.7 into partial fractions we have

$$y(p)_2=\mathcal{L}(\ddot{y}_c)\left\{\frac{\xi\omega_1^2}{(\Omega^2-\omega_1^2)(p^2+\omega_1^2)}+\frac{\xi\omega_1^2}{(\omega_1^2-\Omega^2)(p^2+\Omega^2)}+\frac{1}{p^2+\Omega^2}\right]$$

writing

$$\xi=\frac{\xi\omega_1^2}{\Omega^2-\omega_1^2}=\frac{1}{\frac{1}{\xi}\frac{\omega_2^2}{\omega_1^2}+1-\omega_1^2}$$

$$=\frac{1}{1+\frac{1}{\xi}\frac{\omega_2^2}{\omega_1^2}-\omega_1^2} \tag{B.8}$$

$$y(p)_2=\mathcal{L}(\ddot{y}_c)\left[\frac{\xi}{(p^2+\omega_1^2)}+\frac{(1-\xi)}{(p^2+\Omega^2)}\right] \tag{B.9}$$

Solution for Impact Acceleration

Utilizing the Zirac impulse function

$$\mathcal{L}(\ddot{y}_c)=\Delta v$$

$$\delta_2=\frac{\xi\Delta v}{\omega_1}\sin\omega_1 t+\frac{\Delta v}{\Omega}(1-\xi)\sin\Omega t$$

$$\boxed{\delta_2=\frac{\Delta v}{\omega_1}\left\{\xi\sin\omega_1 t+\frac{\omega_1}{\Omega}(1-\xi)\sin\Omega t\right]} \tag{B.10}$$

Note that when $\omega_1 << \omega_2$

$$\Omega^2\to\omega_2^2 \qquad \xi\to 0 \qquad \delta_2\to\frac{\Delta v}{\omega_2}\sin\omega_2 t$$

which is the correct single degree of freedom solution.

Solution for Long Period Acceleration

If we impose a steady acceleration $\ddot{Y}_c$

$$\mathcal{L}(\ddot{y}_c)=\frac{1}{p}\ddot{Y}_c$$

equation B.9 becomes

$$\boxed{\delta_2 = \frac{\ddot{Y}_c}{\omega_1^2}\left[\xi(1-\cos\omega_1 t) + \frac{\omega_1^2}{\Omega^2}(1-\xi)(1-\cos\Omega t)\right] \qquad \text{(B.11)}}$$

Note that when $\omega_1 << \omega_2$

$$\delta_2 \rightarrow \frac{\ddot{Y}_c}{\omega_2^2}(1-\cos\omega_2 t)$$

which is the correct single degree of freedom solution.

Solution for Sinusoidal Input $\ddot{y}_c = a\cos qt$

$$\mathcal{L}(\ddot{y}_c) = \frac{ap}{p^2+q^2}$$

equation B.9 becomes

$$\boxed{\delta_2 = \frac{a\xi}{(q^2-\omega_1^2)}(\cos qt-\cos\omega_1 t) + \frac{a(1-\xi)(\cos qt-\cos\Omega t)}{(q^2-\Omega^2)} \qquad \text{(B.12)}}$$

Natural frequencies occur when the solution moves to infinity, so that the first natural frequency is

$$q_1 = \omega_1 \qquad \text{(B.13)}$$

and the second

$$q_2 = \Omega = \sqrt{\omega_2^2+\xi\omega_1^2(1-\omega_1^2)} \qquad \text{(B.14)}$$

$$= \omega_2\sqrt{1+\xi\frac{\omega_1^2}{\omega_2^2}(1-\omega_1^2)} \qquad \text{(B.14a)}$$

12857

A RESEARCH PROGRAM TO DEVELOP A 60 "G" PERSONNEL RESTRAINT SYSTEM

Howard E. Freeman
Lead Engineer, Vought Astronautics Division
Chance Vought Corporation

Introduction

In 1959 the Protective Equipment Section of the Aeronautical System's Division at Wright Field initiated a program to determine what could be done in the areas of crew restraint and support to permit higher, omnidirectional impact stresses to be imposed on the crewman. They requested a new look at aircrew-protection fundamentals. Specifically, the known and anticipated imposed loads were to be established, human capabilities were to be investigated, and any materials or concepts that might be applicable were to be utilized to develop a system that satisfied requirements. No constraints, such as retrofit or designing for a specific current or projected vehicle, were imposed. It was emphasized that the physiological and internal anatomical characteristics of man were to be primary considerations, as opposed to his obvious external configuration.

The above problem statement resulted in Vought Astronautics "Personnel Restraint System" program. This paper summarizes the program.

Organizing the Program

The primary consideration in organizing the program was integrating the disciplines of engineering and medicine to form the strongest possible team to study and resolve the problem. Dr. Schwichtenberg of the Lovelace Foundation and Dr. Gell of Vought Astronautics were consulted on overall medical policy and major problem areas. Medical specialists worked with the engineers at the detail working level throughout the program. This working relationship permitted both engineering and medical considerations to influence all phases of concept development, selection, and design.

Established Requirements

Representative current and proposed flight systems were studied to determine the types and magnitudes of acceleration stresses that would be generated. The studies indicated a number of significant stresses would be encountered. Of these, omnidirectional impact stresses, with a low total energy and high-peak magnitude, appeared to be one of the most severe and least understood stress areas. Peak "G"'s of 60 "G" or higher appeared probable.

Figure 1 summarizes the protection goals established for this program. The presently accepted HIAD limits[1] for longitudinal transverse accelerations are shown by the heavy black lines. The transverse impact accelerations, with a 60-"G" peak and a total velocity change of 30 feet per second, established as a goal for this program,

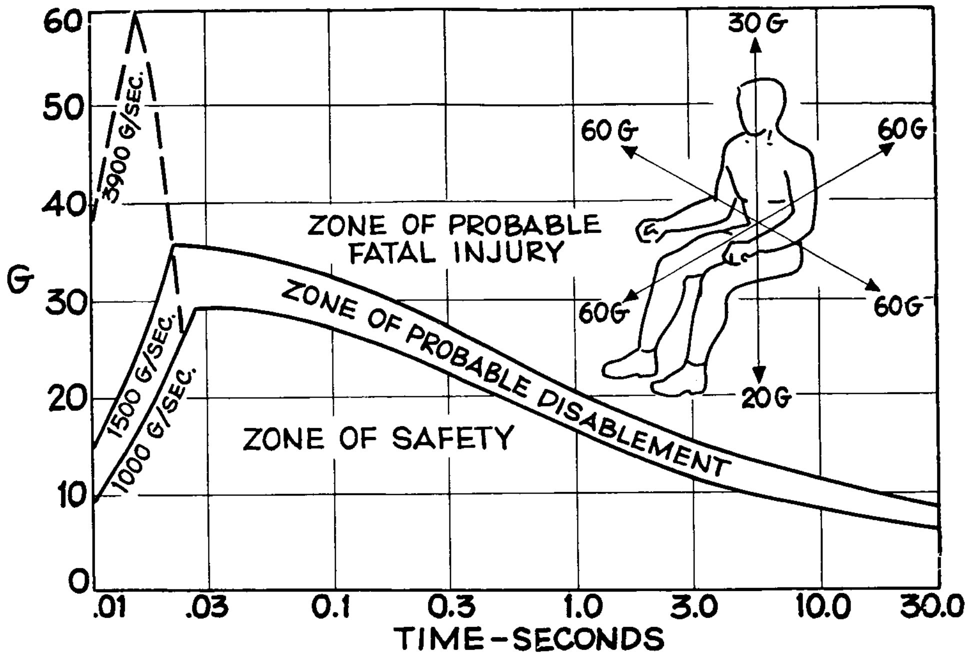

Figure 1. Protection goals.

is illustrated by the dashed line. Protection goals for the six body axes are illustrated on the right side of the figure.

An inertial load analysis was conducted to determine the total force and unit loading that must be imposed to support each element of the body at the established "G" level. The unit load analysis is illustrated in Figure 2. Utilization of most of the

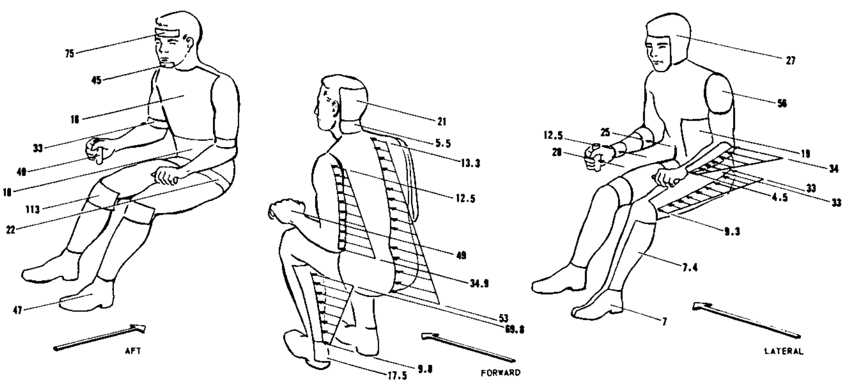

Figure 2. Unit load analysis.

element projected area was assumed in computing the unit loading. For unit weights for the man see references 2 and 3.

Unit loads on the body elements did not appear to be excessive if broad support was provided. Similar unit loads have been safely applied many times on 30- to 40-"G" tests using conventional harness.

Man's internal structure was investigated to determine possible load paths for transmitting the body-element inertial loads.

The location and support mechanism of internal organs was studied to avoid applying external loads that would be directed into an internal organ and to estimate the damage that might result from the organ's own inertia. Engineers on the program participated with medical team members in a canine vivisection and witnessed an autopsy to acquire a better understanding of human internal structure.

Internal organs, with the exception of the heart, are well supported for most acceleration directions if the abdominal wall is not permitted to distend. It did not appear feasible to prevent the heart from moving in the chest cavity; however, if the normal chest contour can be maintained, squeezing of the heart by the rib cage will be prevented.

It was concluded from the human characteristics study that distributing body support over a broad area and maintaining the normal body contour would provide the optimum restraint that can be applied externally.

Developing Concepts

Utilizing the established loads and human structural characteristics, a large number of materials and basic concepts were studied for possible solutions to the restraint-support problem. From these materials and concepts five complete systems were developed to the preliminary design stage and tradeoff studied. The five systems embraced a range of basic concepts from stowed systems that left the crewman unencumbered during normal flight, through strap-net-cloth harness to rigid encasement in contoured shells.

Selecting the System

A team of crew-equipment designers, medical specialists, and experimental pilots evaluated the concepts. Elements of the systems were evaluated separately to assure consideration of all concepts generated. Primary emphasis was placed on the protection afforded by the concept.

A complete system concept was assembled from the mutually compatible elements that had ranked high in the evaluation. This overall concept was detail-designed into a research restraint-support system for manned testing. Figures 3 and 4 are photographs of the system. The central components, illustrated in both figures, are an individually fitted, fiberglass, torso garment and a similarly fitted seat pan. These rigid components were selected to provide broad support and preserve the normal body shape under inertial loading. A flexible, low-rebound liner is used for comfort and intimate fit. The torso shell is retained to the seat structure with steel cables to minimize stretching and the resultant rebound.

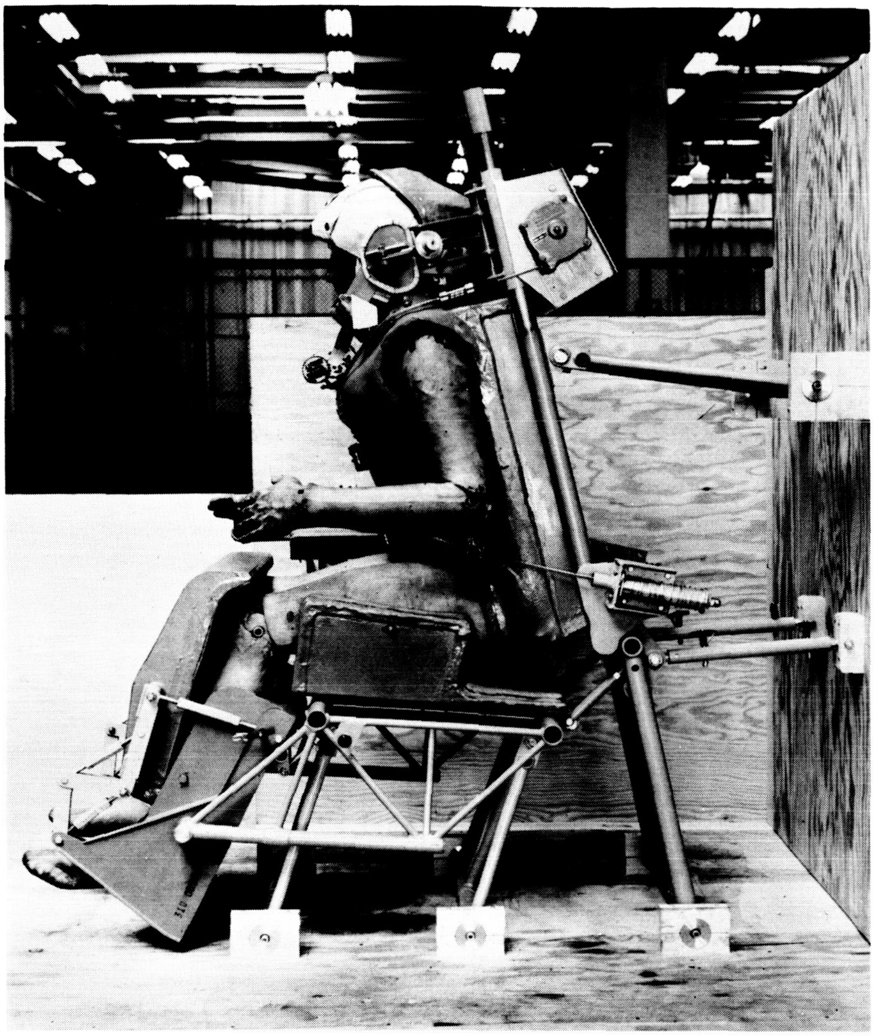

Figure 3. Top view of restraint system.

Head restraint is another departure from current practice. The conventional hard helmet and visor were developed for protection against ejection windblast and impingement on foreign objects. With a shirtsleeve environment and head restraint neither requirement is valid, and the inertial load imposed on the spine by a hard helmet could not be justified. A dacron strap system, positioned by a leather helmet, was chosen to minimize forward head motion. Lateral head supports are mounted on a carriage that adjusts vertically relative to the seat structure for crew-size variations. Low-rebound padding in the helmet cushions the ear area.

Arm support is provided by contoured armrests and hand-holds with a strap passed over the crook of the arm holding the arm back and down.

The subjects legs are positioned and restrained by the sides of the seat shell, a central divider, a contoured leg backrest, and a leg cover. Antisubmarining protection for the torso is also provided by the leg cover, which supports the forward inertial loads of the thighs and legs and stabilizes the pelvis by a direct load path through the

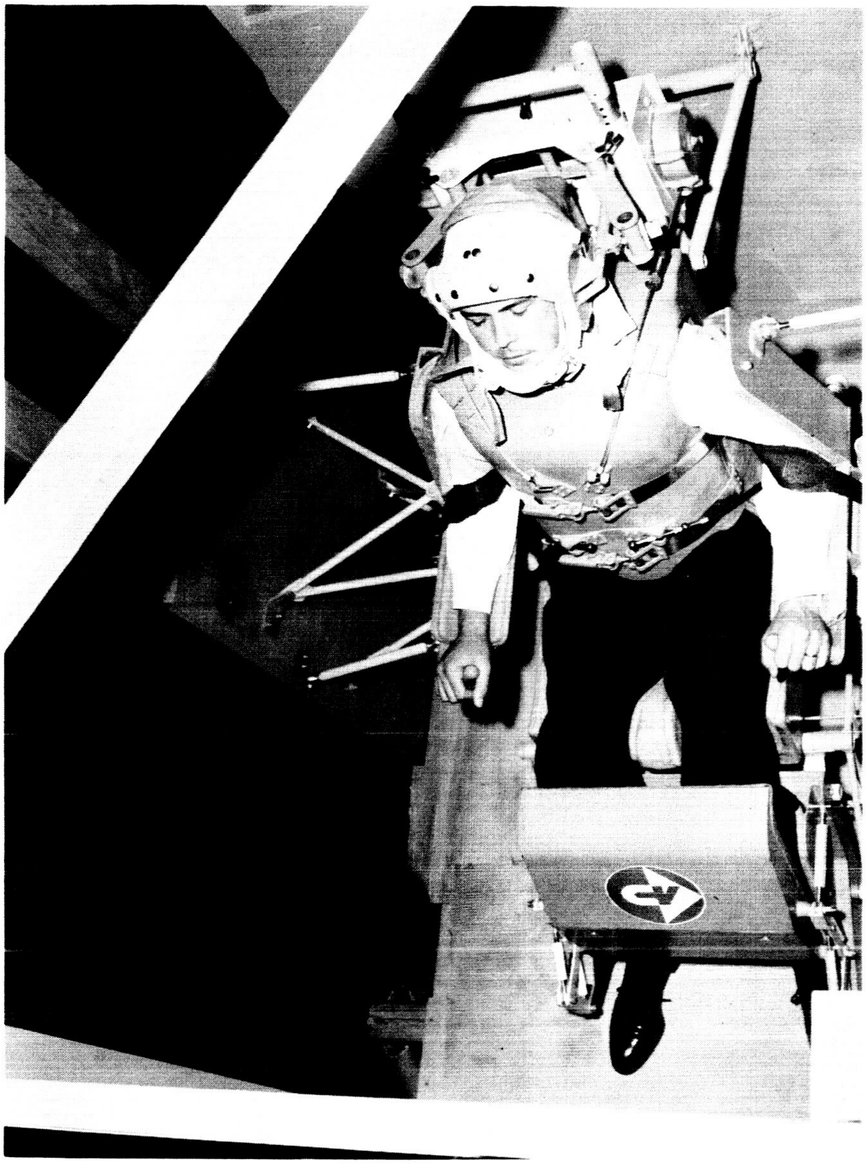

Figure 4. Side view of anthropomorphic dummy in full restraint position.

femur into the pelvic socket. The contoured lower skirt on the torso backshell and the sides of the seat pan reinforce the pelvic socket by limiting lateral shifting of the thighs.

The support structure is a tubular-steel frame articulated to provide a torso-forward position for boost and torso-aft position for less stressful flight elements. Full immobilization and restraint are applied in the forward position. In the next phase of the program a system will be designed to mechanically position and pretension restraint during the transition from the torso-aft to the torso-forward position.

Figure 4 shows the system mounted on an ASD mockup fixture. The subject, a 95th-percentile anthropomorphic dummy, is in the full-restraint position.

The Aeronautical Systems Division is currently testing the system, with the dummy in a six-directional drop-test fixture. When the structural integrity and mechanical characteristics of the system have been established, manned testing will commence.

Concluding Remarks

It should be re-emphasized, in conclusion, that this system is intended solely as a research tool, directed toward impact-testing of live subjects. The forthcoming tests should yield valuable data on the characteristics of "rigid" restraint and the subjects inherent tolerance to acceleration stress when broadly and rigidly restrained.

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FULL BODY SUPPORT SYSTEMS

R. F. Gray
Aviation Medical Acceleration Laboratory
U. S. Naval Air Development Center
Johnsville, Pennsylvania

The primary damaging effect of acceleration is the distortion of the body. To the extent that such distortion can be prevented, persons will be protected from acceleration. Since the problems are basically those of mechanics, the solutions are basically solutions to mechanical problems. The author has chosen as an area for research and development of protective systems those acceleration levels which cannot be tolerated by the action of physiological mechanisms only, but which require artificial aids.

The most challenging problem has been that of increasing tolerance to positive G; that is, increasing tolerance to acceleration tending to stop circulation to the head because of forces tending to keep blood in or force it to the lower part of the body. A major portion of the challenge seems to lie in supporting the circulatory system within the chest. As a background, World War II studies had indicated that the reason for loss of consciousness during positive acceleration ($+A_z$) was that blood tended to pool in the body below the heart and was not available for the heart to pump to the head. These studies also indicated that if pressure as a function of the acceleration level was applied to larger and larger areas of the body, tolerance would increase up to a limit of about three G above the normal acceleration tolerance of a given individual.(1) This was true when pressure was applied to all of the body below the chest in humans and to chin level in animals. Jasper(2), in his unpublished report, "Centrifuge Experiments with Animals," noted that in animals protected by submersion in water, death seemed to be associated with dilation of the heart. Jasper's comments along with the data suggested that, as a first goal, it would seem desirable to limit the maximum possible amount of blood in the chest to that which was present in the respiratory system at the start of the period of acceleleration. This would, in turn, limit the amount of blood available for distending the heart through the action of acceleration. Such limitation of blood volume was to be accomplished by means of air pressure within the chest. It would be necessary to protect the chest, the circulation, and areas of the body such as the abdomen against the effects of the pressure. Experiments on mechanical models further indicated that such a pressure system might not only limit the amount of blood in the chest but might also maintain blood in the arteries to the head and maintain the integrity of the blood circuit between the heart and the head. If this held true, then circulation to the head might be maintained at higher positive-G-levels than attainable at that time.

Attempts at animal studies indicated that a proper study of these effects could not be conducted with animals and it was decided to conduct studies with humans. The major hazard to be faced seemed to be that of aeroembolism. Such aeroembolism had been observed at the New London Submarine Base Laboratory to be associated with overexpansion of the chest as submerged subjects floated upward in water while retaining air in their chests which had been breathed under pressure while submerged. New

London found that taping of the chest gave dogs complete protection against aeroembolism.[3] Also, Dr. Karl Schaefer at New London showed the author newly obtained records indicating pressure differences between the mouth and the water which regularly reached levels of 10 psi to 15 psi in subjects rising or descending in water. The lower pressure was within the mouth during descent. Presumably, these measurements also represented the pressure differences between the mouth and the lungs.

From these data it was decided that a reasonably safe, simple, and adequate test of the hypothesis of increased positive G-tolerance through the addition of internal chest pressurization might be made by simply seating the subjects in water to eye level and having them hold their breaths during periods of centrifugation. Two subjects sustained levels of 10 and 10.5 G respectively without experiencing greyout or reaching tolerance limits. A third subject achieved a level of 16 G without dimming of the visual field. Air forced from his chest led to a stretching of the cheeks and other facial tissues. Air forcing its way out of his mouth or out through his nose irritated the pharynx and precluded attempts to attain higher G-levels. The record of this run is shown in Figure 1 and is compared to a run made by this subject in the same tank without water in the tank. This subject's increase in tolerance was 13 G or more than four times the increase in tolerance previously gained with any body support system. The goal of the program had definitely been achieved. It was shown that support of the interior of the body could be reasonably achieved and that this support could be of significant value. The goal of gaining increased support for structures within and around the gas-filled spaces of the body still seems the major area of hope for improving tolerance to linear accelerations.

Much useful work can be done, however, from a short-term point of view in improving tolerance to linear accelerations by improving supports for the outside of the body. Shields to keep sharp points from striking the body with concentrated force or restraints to keep people from hitting sharp corners specifically illustrate the nature of some problems to be solved by external support systems. Head rests to prevent the whiplash injury to the neck illustrate a third function of the external support systems in minimizing those differential inertial effects known as flailing. The fourth function of external support systems is well illustrated by the G-suit which keeps certain regions of the body from expanding and thus opposes the entry of blood into the region.

So far as is known, the maximum external support to the body utilized up to the present has been in the G-capsule (Figure 2) at the Aviation Medical Acceleration Laboratory, Johnsville, Pa. In use, water was brought up to chin level on any subject in the device. The subject, wearing a swimming suit, oral-nasal mask, special eye glasses, and finger switches, would then rub bubbles off himself and the sides of the capsule and shake the water vigorously and generally do his best to get air out of the water. Thereafter, more water was brought into the capsule until it ran out of the standpipe above the subject. This standpipe was closed and again the subject removed as many bubbles from the water as he could. These bubbles would collect in the form of a small gas pocket above the subject's head. Thereafter, more water would be brought into the capsule and out through the standpipe above the subject's head. The capsule would then be adjusted and the experiment would begin. It seems that very little more could have been done to achieve full external support. The eyes, for example, were directly submerged in water and especially designed eyeglasses compensated for loss of corneal focusing effects. The only area of possible improvement in this respect seems to be that of reducing the size of the oral-nasal mask cavity.

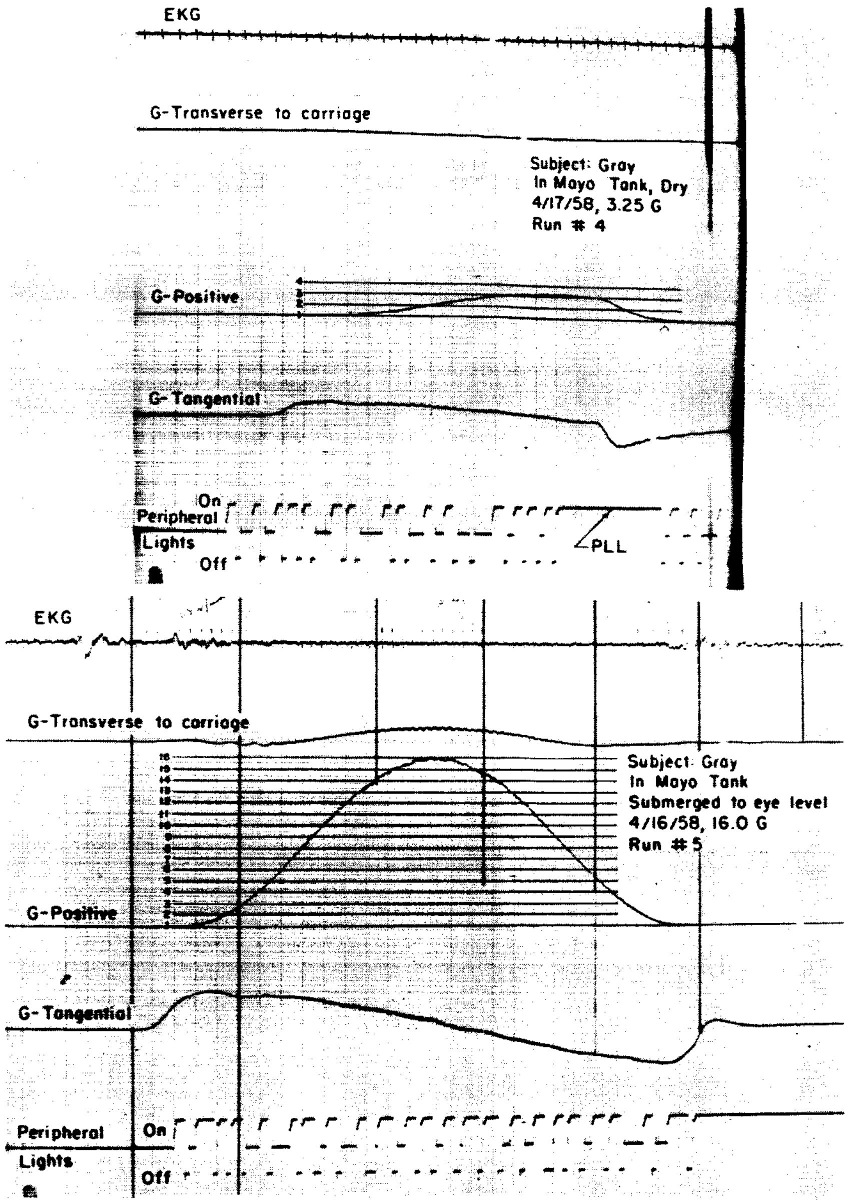

Figure 1

The skin-diver type of mouthpiece was not used because it was felt that it might be ea[illegible] lost by the subjects and thus increase the hazard of drowning.

Subjects in this capsule were exposed to centrifugation while facing away from the center of rotation. Three subjects have been exposed to maximum levels of up to 26 G, 28 G, and 31 G, respectively, during acceleration periods of 25-30 seconds total time, up to a maximum acceleration level and back down. Slight nosebleeds occurred for each of the three subjects following exposure at levels of 23 G, 28 G, and 31 G, respectively. Frontal sinus-type pain was first reported by each of these subjects at these G-levels. Abdominal pains occurred for all three subjects at levels above 10 G, but these pains could be almost eliminated through straining of the abdominal

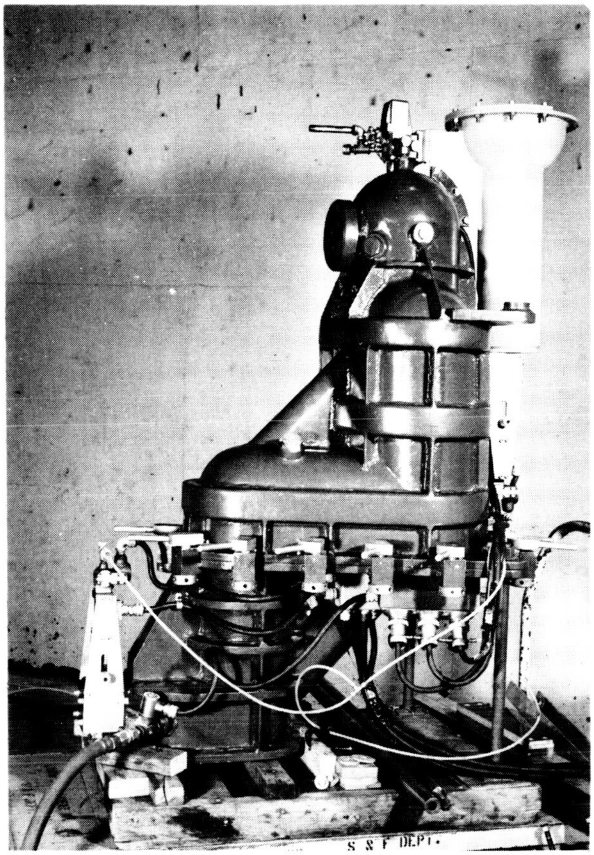

Figure 2

musculature. Previous centrifuge experience had indicated tolerance limits in this position of between 10 G and 15 G for different individuals. Limits of tolerance were defined by eye pain or pain in the legs. Neither of these effects occurred during studies in the G-capsule. It therefore seems that, indeed, full body-support systems can give worthwhile increases in G-tolerance.

Submersion in water, while it does give a nearly perfectly fitting support system and aids movement within the container during periods of acceleration, has the disadvantage of great weight and the hazard of drowning, and sets up large uncompensated gradients of pressure around the gas-filled regions of the body. If a cast were formed around the body, the cast would have many of the attributes of water as a support system. A lightweight material which could be used to form this kind of cast is foamed polyurethane. As used in experiments at Johnsville, this weighed between 1.5 and three pounds per cubic foot, or had between 1/20th to 1/40th the density of water. In small volumes, temperatures, due to the exothermic type of reaction forming the foam, went up to about 180°F. However, when cast into large blocks completely submerging a human subject, temperatures in some regions of the foam were measured at about 270°F. While these temperatures sound high, the low heat content per unit volume of the material, coupled with its excellent characteristics as an insulator, allowed people to tolerate periods of up to 20 minutes of complete encapsulation without cooling procedures. Various parts of the body had to be protected with between one and four layers of cloth (cotton or knitted wool). After 20 minutes, cooling by means of ventilated underwear allowed the subjects to stay in the foam for as long as two hours. Polyurethane is toxic in the sense that its isocyanate compounds are irritants of the upper respiratory system, eyes, mucous membranes, and skin. Direct contact must be avoided, the face protected by a mask, and breathing air must come from a source not contaminated by isocyanates.

The particular polyurethanes utilized in the studies at Johnsville (Thiokol, Rigidthane 334) became quite rigid in about 30 seconds after pouring. This rate of hardening was quite satisfactory as the foam pressed tightly against stationary parts of the body, but respiratory excursions left sufficient space around the trunk to permit normal breathing with limited maximal inspirations. The expansion of the foam developed pressure sufficient to produce pain in the head of one subject after 15 minutes. A knitted hood gave two subsequent subjects adequate protection against the pressure.

Of the various possible methods of forming the foam around the subjects, placing the subject in a containing box (Figure 3) and filling the space between the container and

the subject with foam formed in plastic bags (Figure 4) proved to be the most convenient. To remove the subject from the foam, the sides of the box were separated at the corners and the bags of foam lifted from around the subject (Figure 5). This was done in two minutes. Very few trials were necessary to determine bag configurations and locations to assure good body support and to minimize bridging of the foam from one part of the body to another. Only very small regions of the body were unsupported.

Figure 3

Figure 4

When this development was completed, a subject was submerged in the foam and exposed to lateral acceleration along the Y axis. He judged the foam to give better support against lateral acceleration than that offered by rigid supports or strap supports, in the sense that it allowed less body displacement. This experiment was terminated because the subject developed severe leg pains, probably as a function of the long period of immobilization.

From the foregoing discussion it can be seen that, while the foam has weight and safety advantages compared to water, it does not fit the body so well, cannot be used to support regions such as the eyes, and a person within it is severely immobilized. Foam castings can give much more complete support than strap systems and can be rapidly formed around or rapidly removed from subjects. Foams seem best applied to one-shot short-term (15-minute) support applications, but some reversible foam-casting support systems have been hypothesized.

Figure 5

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AN ANALOG COMPUTER WHICH DETERMINES HUMAN TOLERANCE TO ACCELERATION

Peter R. Payne
Frost Engineering Development Corporation

Introduction

Although the concept of a mechanical analog of man was conceived independently by both Kornhauser and the author, it goes back much further than their publications. This is not particularly surprising of course; probably almost any mathematically-oriented engineer with dynamics experience would decide upon the same initial approach, since it has been a commonplace in other branches of engineering for many years. Indeed the first German investigators to obtain information on the breaking loads of spinal vertebrae(6,7) used lumped parameter dynamic models, almost as a matter of course, in order to predict tolerable acceleration limits in the spinal direction.

It is ironical to note that in this field, as in so many others, the war-time German workers were many years ahead of their Allied counterparts, and that their leadership was not based on lavishly funded research programs, but on a clear insight into the mechanisms of the problem they were investigating, reinforced by a sound academic background.

Among the first notable additions to this early German work was that contributed by Latham(8), who extended the theoretical work to include an additional degree of freedom to represent the restraint system, and utilized analog computer techniques to determine the effect of varying the dynamic characteristics of an ejection seat cushion.

In this country a number of investigators(9-13) have attacked the problem along similar lines since 1946, but theirs seem to have been voices crying in the wilderness, and their work was essentially ignored by the engineers concerned with hardware and, indeed, by some of the research workers who followed them. There must have been a number of reasons for this unfortunate state of affairs, but the lack of any continuing support by the appropriate governmental agencies concerned was certainly a major cause. Also, of course, most of the programs were directed and staffed by medical personnel and/or mechanical designers, who presumably were unaware of the contributions that the dynamics research could have made to their work—one more unhappy example of the waste caused by lack of communication between disciplines.

It is an obvious truism to say that significant and substantial progress in a new technology can be achieved only by a continuing research program, but it is also one which past events have shown will bear repetition, particularly in a field such as "Body Dynamics," where there is little in the way of end-products from which a commercial profit can be made. Thus, research funding of any significant amount can be expected only from governmental agencies.

In 1961 one of the first evidences of recognition of these facts came when NASA supported work initiated by this author(14). With this precedent established (and gratefully acknowledged), and because of the importance of body dynamics research to programs such as Gemini and Apollo, it can be hoped that support of this type will now be continuous.

A Survey of the Development of the Theory of "Body Dynamics"

The human body is capable of withstanding relatively large accelerations, provided they endure for only a short period of time. For durations of less than one second the limitations are mainly structural, and the human body can be regarded as a non-linear, visco-elastic, distributed mass dynamic system, subject only to the laws of mechanics.

The need for increased accuracy in predicting the maximum acceleration levels that can be tolerated has led to the development of a new technology—"body dynamics." It has achieved prominence rapidly, because it makes possible substantial improvements in the overall performance of manned aerospace systems by providing a scientific basis for reduction of weight and simplification of certain systems.

Perhaps the most common example of this is the ejection seat, where the limitation of human tolerance constitutes by far the most important design criterion and is the greatest handicap to substantial reductions in seat weight and cost.

For simple trapezoidal acceleration-time histories it has proved possible to draw tentative limits for tolerable acceleration, based primarily on the test results obtained by pioneering experimental workers. Unfortunately the experimental points available are so few, and the margin for alternative interpretations is so wide, that each investigator seems to have drawn an entirely different "limit" envelope through the same experimental points. Some of these "gross body limits" are shown in Figures 1 and 2 for the case of a seated man subjected to spinal and transverse accelerations.

The first major improvement in the accuracy of the "limit" definition became possible when it was realized that the human body is still subject to the mechanical laws which govern the rest of the physical world, and that when subjected to acceleration it must behave in accordance with the laws of mechanics. In other words, it must behave like some sort of dynamic system.

The next step was to examine mathematically all the possible behavior modes of the types of dynamic systems, both linear and non-linear, which could reasonably be expected to be analogous to the human body. Because a precisely similar problem has not arisen previously in applied mathematics, it is likely that this theoretical investigation will continue for some years before our knowledge is completely adequate. Nevertheless, several highly important facts have already emerged from this study. For rectangular acceleration-time inputs of the type shown in Figure 3, for example, it has been shown(1) that <u>any</u> dynamic system will behave in a generally similar manner. If we limit the load or stress in the system to some arbitrary figure, then the acceleration necessary to impose this critical load will vary with the duration over which it is imposed.

For very short durations we find that, for all linear and non-linear systems so far studied, the velocity change represented by an acceleration pulse is the only significant parameter. For longer durations only the acceleration is important, the

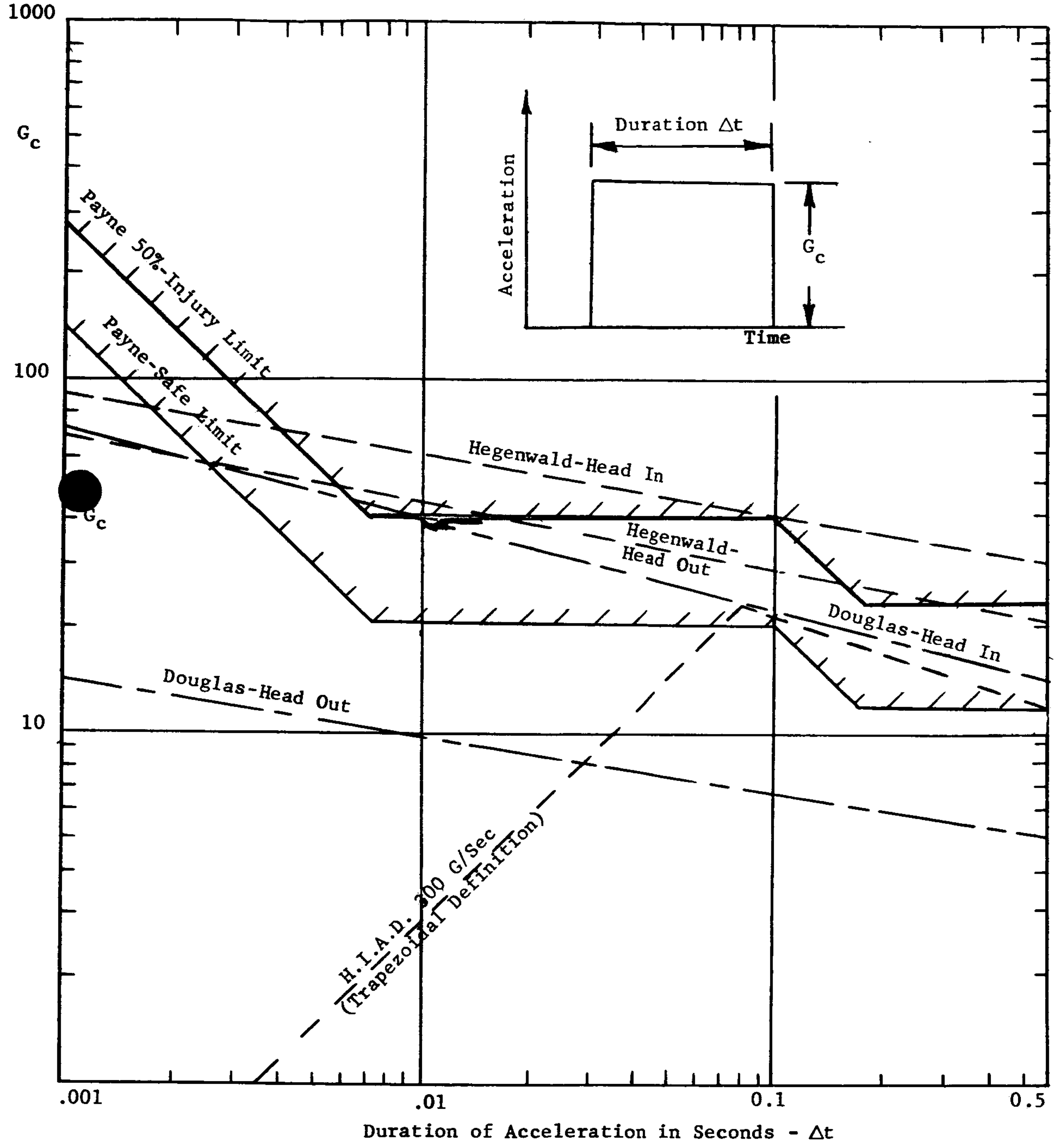


Figure 1. Human Tolerance to Spinal Acceleration (Seated Man)—Gross-Body Limits

Note that the HIAD curve is based upon a trapezoidal definition, and is therefore more severe than appears from this plot. The concept of a "gross-body" limit implies that the head restraint is "soft."

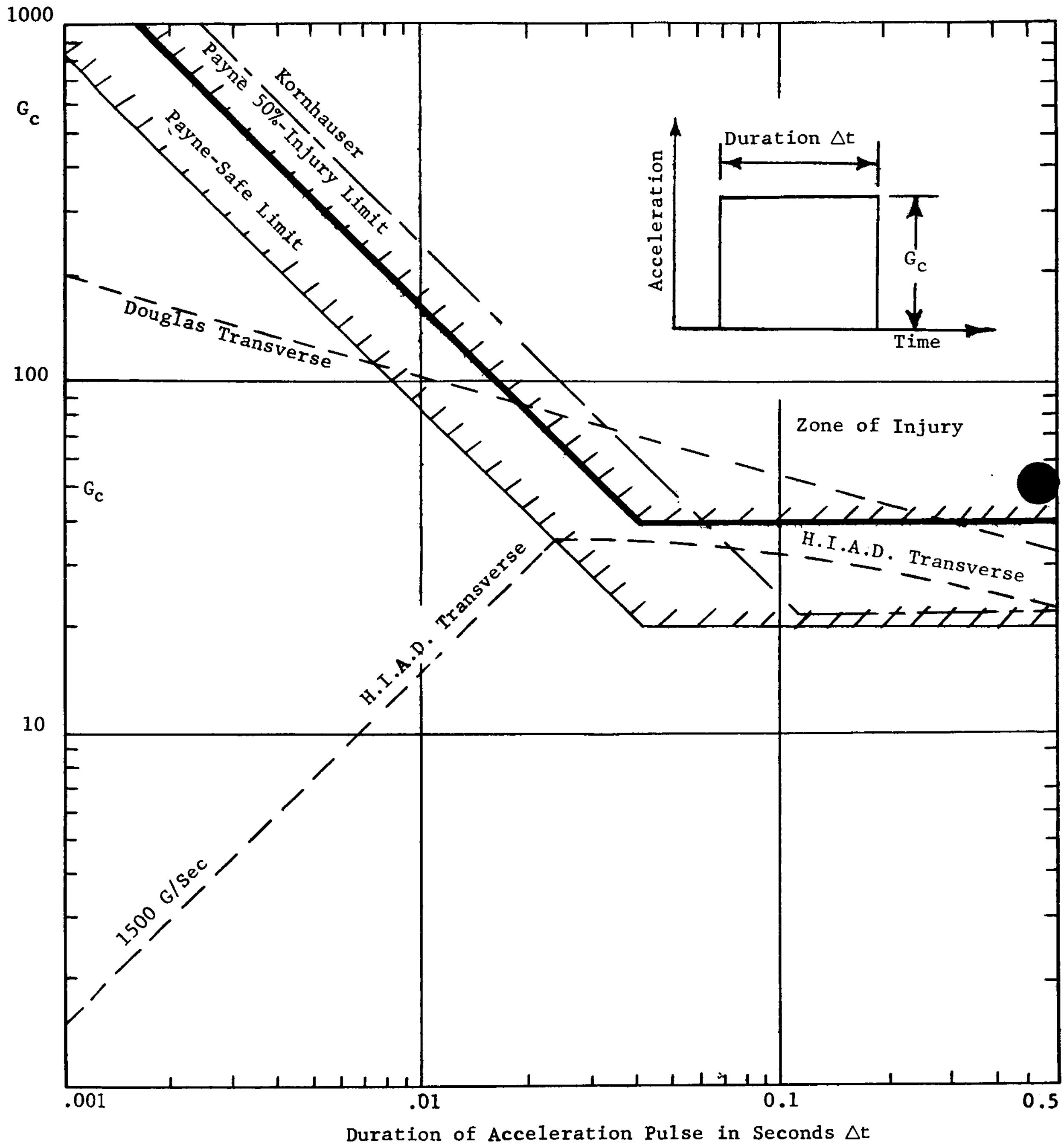


Figure 2. Human Tolerance to Transverse Acceleration (Seated Man)—Gross-Body Limits

Note that the HIAD curve is based upon a trapezoidal definition, and is therefore more severe than appears from this plot. The concept of a "gross-body" limit implies that the head restraint is "soft."

velocity change which it represents having no influence whatsoever. This knowledge can be summed up in the tolerance graph sketched in Figure 4, which is generally applicable to any type of system with one degree of freedom. When a system has several degrees of freedom, a series of tolerance curves of the type shown in Figure 4 are overlapped, although not by elementary superpositioning.

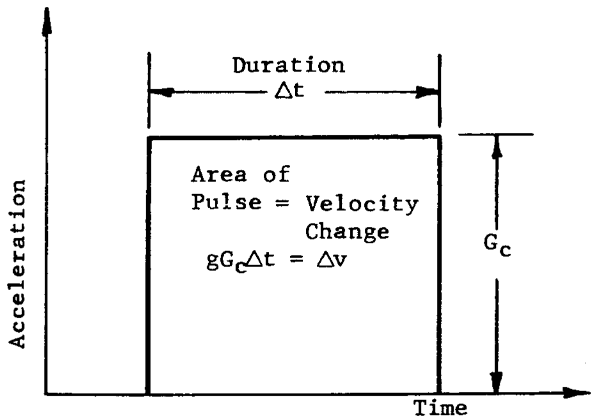


Figure 3. Idealized "Square Pulse" Acceleration

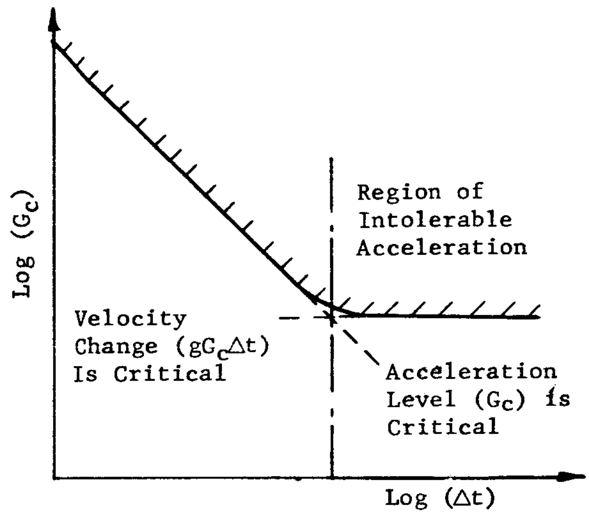


Figure 4. Tolerance Graph for a Single-Degree-of-Freedom System

With this information we can draw a much more accurate "limit envelope" through our scanty experimental data for the limit of human tolerance, since we know that the left-hand side must be a line at 45° to the axes (for an equal log plot) and that on the right side of the graph, the line must be horizontal. The picture is a little more complicated with a multi-degree-of-freedom system, but the same general principles still apply. Obviously we have more closely defined where the limit line must fall when we have defined the slope with which it must pass through the experimental points; moreover the use of this concept enables us to use "impact data" which previously were unusable, thus increasing the available data points, and also to use (with caution) the results of impedance data obtained by subjecting humans to sinusoidal vibration.

So far we have discussed only the effect of a rectangular acceleration pulse of the type shown in Figure 3, and its concomitant limit case of an "impact acceleration" in which the acceleration is very large but endures only for a very short period of time, that is, a duration which is an order of magnitude less than the natural period of the dynamic system upon which it is imposed. We have seen that a very simple hypothesis, in which the human body is represented by a single mass spring and damper system, adequately predicts the available experimental data. The first investigator to point out this agreement was Kornhauser[4] who confined his attention to an undamped, linear, single-degree-of-freedom system and demonstrated its agreement with the available experimental data. In later papers[2, 5] Kornhauser discussed this approach in more detail, and reported an extensive test program with mice. This program demonstrated the validity of his thesis, within the limitations of the acceleration recording equipment employed on his test rig.

In subsequent work[15] the writer showed that any single-degree-of-freedom system is sensitive only to velocity change when the acceleration duration is small, and

that this law holds true for non-linear as well as linear damped systems. Thus any single-degree-of-freedom system that we can conceive will give a tolerance graph of the type shown in Figure 4.

In practical aerospace vehicles and escape systems, the acceleration-time history bears little relationship to the idealized rectangular pulse discussed hitherto in this section, and examined (with variations) by the early workers in this field. Thus the work so far described is of little value in normal engineering applications, although it does increase our knowledge of the fundamentals of "body dynamics." It will be recalled however, that the tolerance graph in Figure 4 was drawn, for a mathematical model, by assuming that there was a critical load in the spring which represented the limits of tolerability. In physical terms we may say that this load in the spring is analogous to the compressive load in a man's spine, for the case of a seated man subject to spinal acceleration, and that the critical load is the value at which a compressive failure of one element of his backbone occurs. It is unnecessary to know the location and mechanism of a failure in the human body however, but only the acceleration-time history which causes the failure to occur. Then by calculating the corresponding load in the spring of a hypothetical model we can draw a suitable tolerance line.

The use of a dynamic model to assess the tolerability of an arbitrary and irregular acceleration is now obvious. By solving the equation of motion for the dynamic system when subjected to the arbitrary acceleration being assessed, we can determine the peak force in the spring. This solution can be obtained either numerically, by hand or on a digital computer, or by use of an analog computer. If the peak force developed in the spring is greater than the value previously determined as the limit of tolerability from experimental data with human subjects subjected to rectangular pulse inputs, then the acceleration is presumed to be intolerable. Conversely, if the peak spring force is less than the critical value, the acceleration is tolerable.

For convenience the spring force is expressed as a "Physiological Index" (P.I.) which is directly proportional to force, but which has g (acceleration) units. Thus when the man is subjected to a steady acceleration of one g, his P.I. is also one g.

Until recently the P.I. was calculated on the assumption that the human body is a linear dynamic system. So long as we are concerned only with the limits of human tolerance, the peak deflection in all cases is of about the same value, and any attempt to introduce non-linear dynamics would have been a needless complication, even if there were sufficient data available to define the precise nature of its non-linearity with any accuracy. For the transverse acceleration case, a single-degree-of-freedom representation was found to be adequate, but for the spinal case, when the man is seated, a two-degree-of-freedom model is required for the theoretical tolerance limits to agree with experimental observations, for accelerations of up to one second in duration.

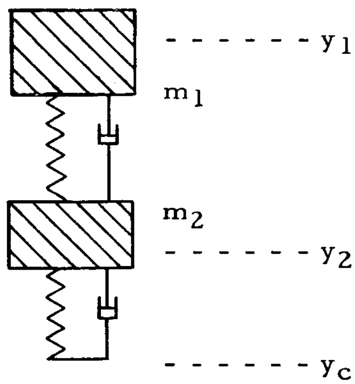
When the duration of the acceleration is less than 0.1 seconds a single-degree-of-freedom model is also adequate for the spinal direction. For longer durations, however, the low frequency visceral mass "deflects" sufficiently to impose additional loads on the spine during the time of application of the acceleration, and the tolerable acceleration level is accordingly reduced.

The linear models so far established for a seated man are illustrated in Figures 5a and 5b which are abstracted from Payne[1]. It is important to note that the models with damping ($\tau = 0.13$) in Figure 5a give the same results as the zero damping models in Figure 5b for trapezoidal acceleration inputs. The model with damping is obviously

SPINAL ACCELERATION - SEATED MAN

"Visceral Mode"

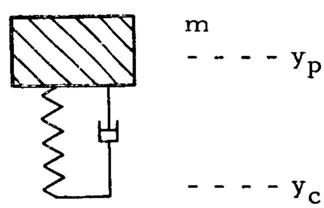
"Spinal Mode"



Ordinates

		"Visceral Mode"	"Spinal Mode"
$\omega_1 = \sqrt{k_1/m_1}$		14.63 rads/sec	---
$\omega_2 = \sqrt{k_2/m_2}$		---	287.5 rads/sec
$m_1/m_2 = 1.368$			
δ_{2CRIT}			0.258 ft.
$\cdot 13(PI)_{CRIT} = \omega_2{}^2\,\delta_{2CRIT} =$			66.2 g
"Corner duration"	$(\Delta t_1 = 2/\omega)$	0.273 secs	.0073 secs
Critical velocity change	$(\Delta t < \Delta t_1)$	126.0 ft/sec	9.27 ft/sec
Critical acceleration	$(\Delta t > \Delta t_1)$	23.0 g	40.0 g

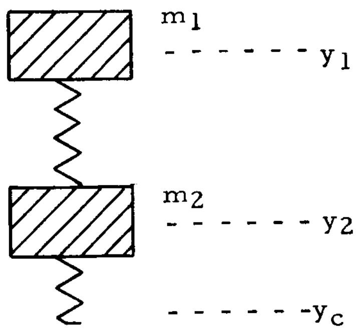
TRANSVERSE ACCELERATION



$\omega = \sqrt{k/m}$		49.6 rads/sec
$\cdot 13(PI)_{CRIT} = \omega^2\,\delta_{CRIT} =$		66.2 g
"Corner duration"	$(\Delta t_1 = 2/\omega)$	.0444 secs
Critical velocity change	$(\Delta t < \Delta t_1)$	53.6 ft/sec
Critical acceleration	$(\Delta t > \Delta t_1)$	40.0 g

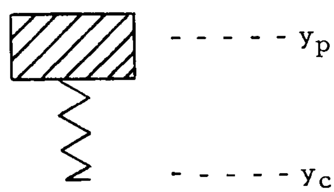
Figure 5a. Linear Dynamic Models (With Damping) of the Human Body Used for Calculating the "Physiological Index" (P.I.) n.b., Models in Figures 5a and 5b yield the same results for trapezoidal input acceleration.

SPINAL ACCELERATION - SEATED MAN



	"Visceral Mode"	"Spinal Mode"
$\omega_1 = \sqrt{k_1/m_1}$	11.8 rads/sec	---
$\omega_1 = \sqrt{k_2/m_2}$	---	278.0 rads/sec
$m_1/m_2 = 1.24$		
δ_{2CRIT}		.0333 ft.
$0^{(PI)}{}_{CRIT} = \omega_2^2\ \delta_{2CRIT}$		80.0 g
"Corner duration" ($\Delta t_1 = 2/\omega$)	0.17 secs	.0072 secs
Critical velocity change ($\Delta t < \Delta t_1$)	126.0 ft/sec	9.27 ft/sec
Critical acceleration ($\Delta t > \Delta t_1$)	23.0 g	40.0 g

TRANSVERSE ACCELERATION



$\omega = \sqrt{k/m}$	48.0 rads/sec
$0^{(PI)}{}_{CRIT} = \omega^2 \delta_{CRIT}$	80.0 g
"Corner duration" ($\Delta t_1 = 2/\omega$)	.0417 secs
Critical velocity change ($\Delta t < \Delta t_1$)	53.6 ft/sec
Critical acceleration ($\Delta t > \Delta t_1$)	40.0 g

Figure 5b. Linear Dynamic Models (Zero Damping) of the Human Body Used for Calculating the "Physiological Index" (P.I.) n.b., Models in both Figures 5a and 5b yield the same results for trapezoidal input acceleration.

more desirable for general work, but in many cases, particularly the analysis of impact problems, the mathematical simplicity of the zero damping model makes its use desirable.

Other models exist for the "head acceleration" limits in Figure 6 and the "standing man" limits in Figure 7, the latter being an extension of the spinal model given in Figure 5a. The "head acceleration" model treats the skull and its contents as an elastic solid, however, rather than as a lumped parameter system, since this appears to be more in accordance with the true physical picture.

It should be noted that the precise values used for the coefficients in these models are subject to change as more experimental data become available, but that we can expect these changes to bring us successively closer to some "correct" value.

The concept of a critical force in the spring, or critical P.I. is of course largely academic, because no two human bodies are precisely alike. The next logical step therefore is to define the limits within which a failure will occur. Once again this is a normal engineering problem and we should expect to find that if we were able to carry out a large number of tests then most failures would occur around some median P.I., but a few will fail at substantially lower and substantially higher values. This is well borne out in practice, and experiments with fresh cadaver vertebrae have shown that the distribution of failure with respect to compressive load is approximately Gaussian in shape and similar to Figure 8.

We should expect to find the same type of distribution, as a function of P.I., for the overall injury level of the human body. There is certainly no possibility of constructing such a statistical variation curve by piecemeal addition of isolated experimental points, such as the experiments with fresh vertebrae, and the most reliable source of data will be experiments with live human subjects.

There are three main sources of such data. In the first place we can draw upon the records obtainable by the Armed Services relative to the effect of short period accelerations imposed upon service men in the course of their duty or in time of war. Arthur Hirsch of the David Taylor Model Basin, U. S. Navy, has proposed to interview crew members of ships which were subjected to underwater explosions during World War II and to follow up the relevant medical records. To within the accuracy with which we can estimate the acceleration-time history to which they were subjected (and applicable data are available at DTMB) we can calculate the P.I. reached by each man. Then from the addition of all the cases of injury available, together with informed estimates of the number of uninjured men at the same locations on the ships, we can determine the probability of injury.

Another source of such operational data is in the results of ejections from aircraft. While accelerating up its rails, an ejection seat imposes large spinal accelerations upon its occupant, and in some particularly bad cases the level of injury is running as high as 50 per cent of all ejections with a particular seat.

Unfortunately, our knowledge of the acceleration-time history imposed upon the seat occupant is not as precise as we could hope to obtain for the case of the ship subjected to underwater explosion. This is because in an ejection seat the man is sitting on a cushion, the dynamic characteristics of which are usually unknown and certainly are never reported in accident statistics, and secondly because the catapult or rocket catapult combination which accelerates him in the spinal direction has an extremely

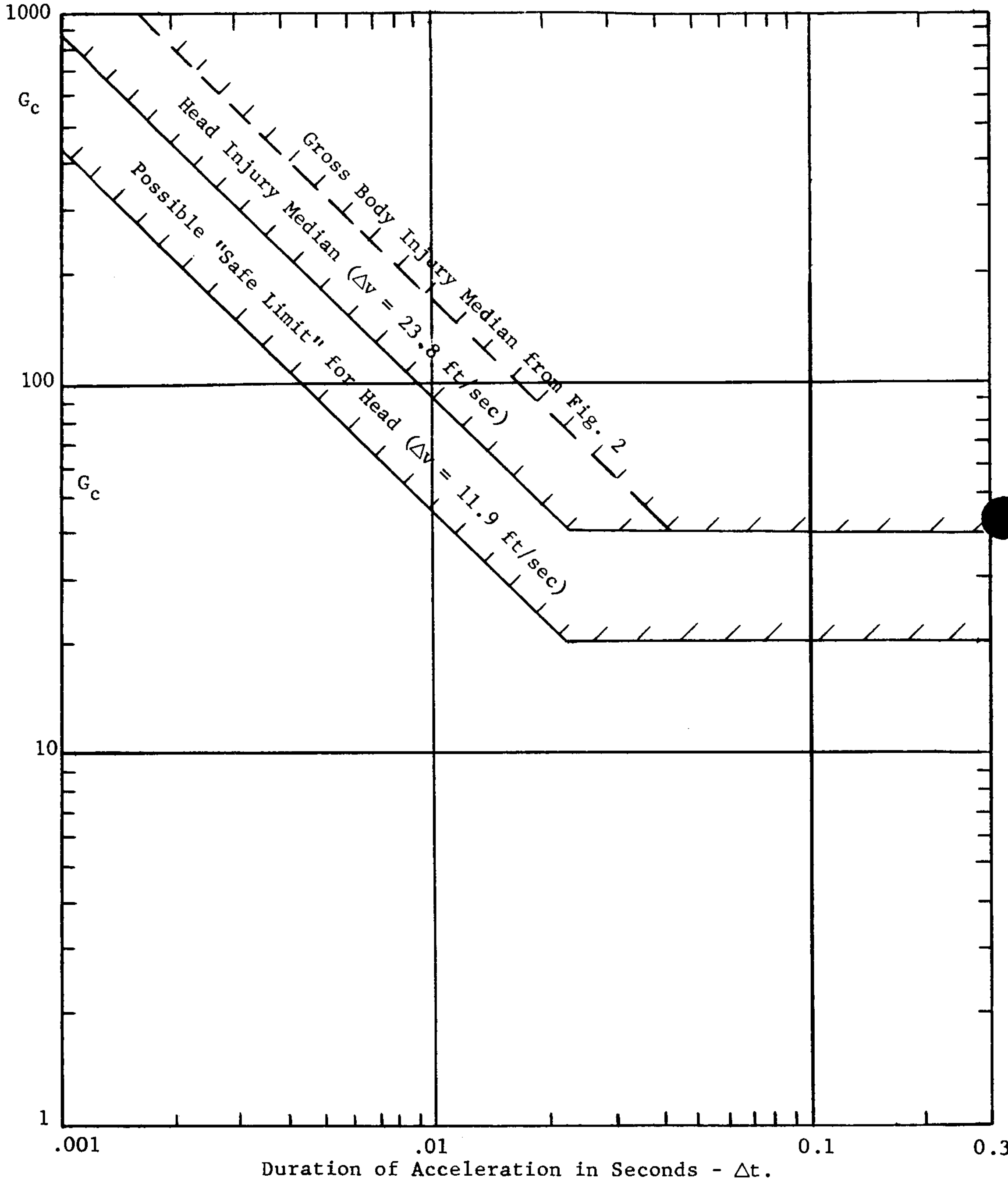


Figure 6. Human Head Tolerance to Transverse Acceleration Based on "Critical Pressure" Model

(Based on the assumption of a firm head support which prevents local deformation of the skull; there is, as yet, insufficient experimental data to permit the discrimination of tolerance levels into the three principal axes.)

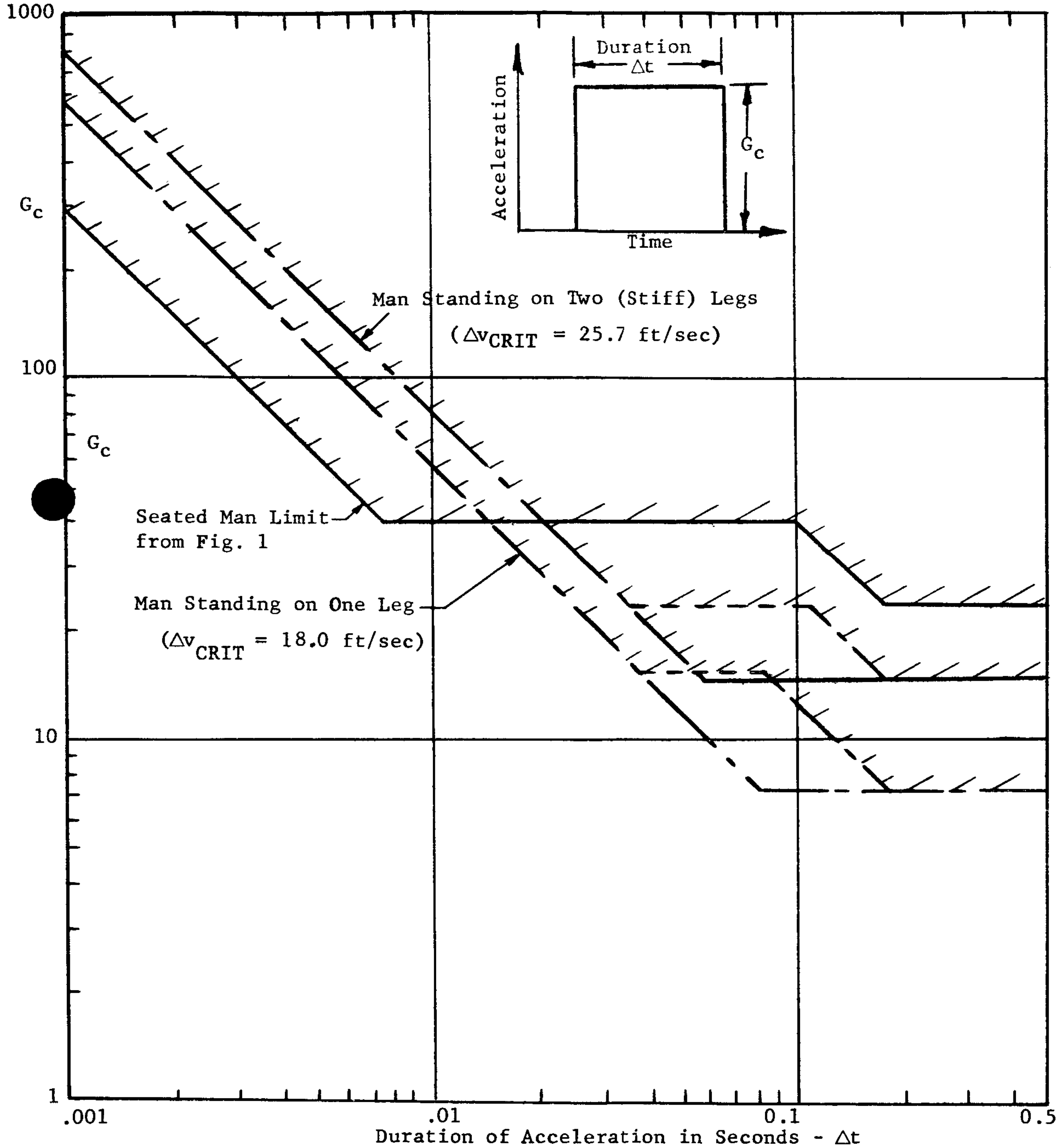


Figure 7. Human Tolerance to Spinal Acceleration as Derived from Experiments with Cadavers. (Standing Man) (Limit Curves are Nominally at the 50% Probability of Injury Level)

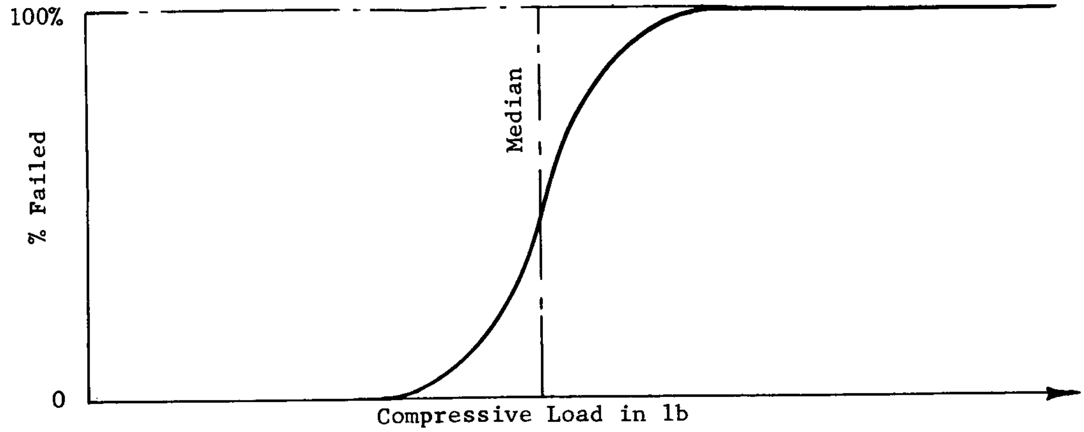


Figure 8. Failure Distribution Curve From Fresh Vertebrae

variable performance. In addition to this, the data currently available do not give the man's weight or state whether he was properly positioned in his seat, both of which factors have a significant effect on whether he will receive an injury. Nevertheless it is possible to draw tentative probability-of-injury curves from the available data, and while these are very approximate, they are certainly an order of magnitude better than any other criterion hitherto available.

Figure 9 summarizes the results of Frost Engineering's analysis of ejection statistics supplied to the company by the U. S. Navy and Air Force, and also by a

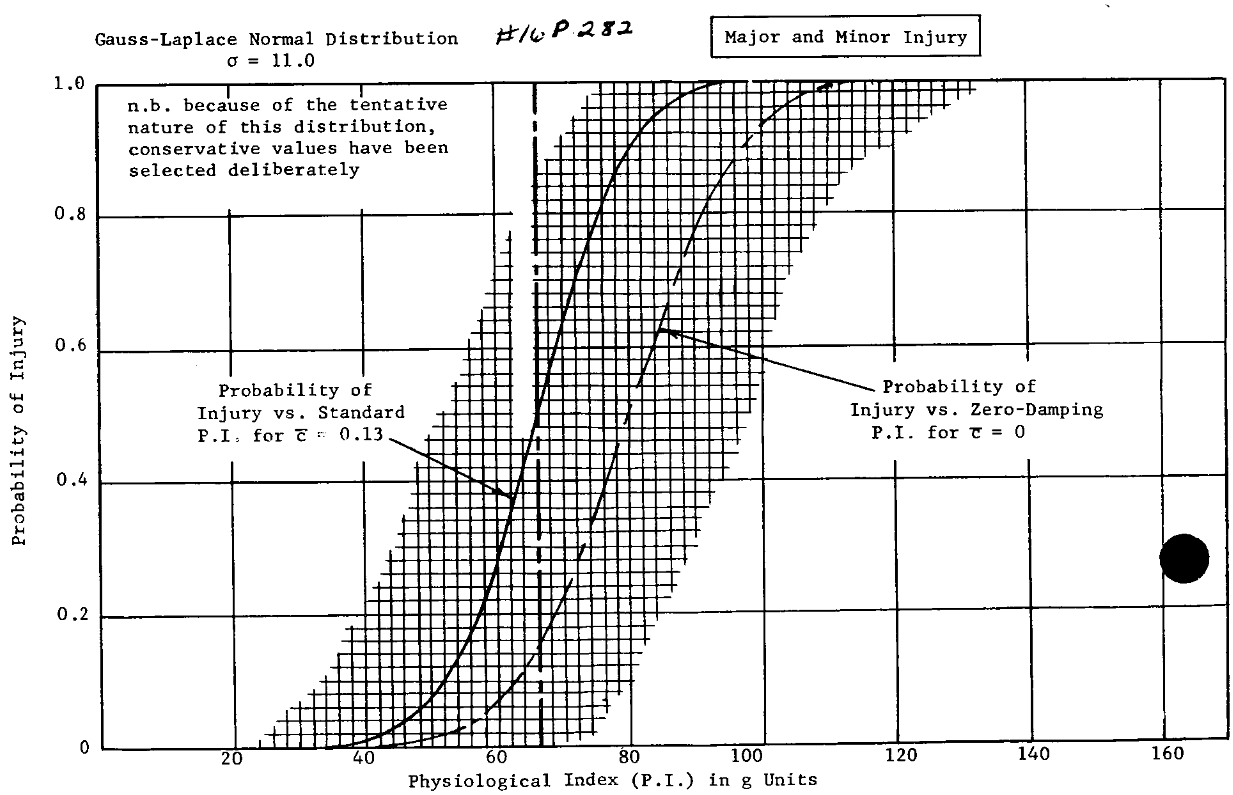


Figure 9. Provisional Probability of Injury Curve for Spinal Acceleration of a Seated Man

number of foreign services. The methods used in this study are described in Payne[3], and it is hoped to accomplish a much more exhaustive study in the future if official assistance and funding can be obtained.

The second source of data for probability-of-injury curves is the experience of the various research agencies that have been conducting physiological experiments. It is known that in many cases such data is relatively useless because the instrumentation used was such that little reliance can be placed upon the input acceleration measurements. Nevertheless there must be many cases in which a short period acceleration was imposed upon human subjects and its magnitude known with reasonable accuracy. It is unnecessary for such data to relate to experiments in which a subjective "end point" was reached, since with our present concept of probability-of-injury, an experiment in which no subjective reaction at all was observed is still an additional piece of evidence that will improve the accuracy of our statistical distribution. Excellent examples of such data are the WADD Aerospace Medical Division experiments reported by Neadley _et al._[16], and the impact experiments of Swearingen[17].

The third source of information lies in tests which have not yet been conducted. By suitably planning future test programs in accordance with established statistical theory, it is possible to improve our estimate of the probability-of-injury, particularly at the low P.I. levels, without reaching an end point in any of the tests. A most powerful tool in this regard is the method of "successive successful tests" which reduces substantially the number of tests needed to reach a given confidence level.

An additional source of information, for certain specific problems, lies in experimental work with cadavers. Although little quantative work has been done with such test subjects in the past, Frost Engineering has carried out some preliminary studies using the excellent work of Draeger, Barr, _et al._[18]

It appears that, because of rigor mortis, cadavers are about twice as "stiff" as live human subjects, so that in an impact type loading they can withstand only two-thirds of the velocity change tolerable to a live subject. For this reason the results of dynamic testing must be treated with caution, and every effort made to correct for the different dynamic characteristics of the cadaver.

For static tests, however, it appears that the results obtained can be used directly, not only to predict statistical variability, but also to give absolute failing loads. For example, the tolerance curve for a standing man given in Figure 7 is based primarily on the cadaver measurements made by Draeger, Barr, _et al._[18], since the "weak link" in this position is known to be the os calcis or the tibia.

Certainly some effort has been wasted in the past because of the lack of a basic theory, and perhaps because research workers using cadavers have not aimed their tests at production of data useful in applied research. It is hoped that the growing acceptance of body dynamics theory will result in more meaningful test programs in the future, and better application of their results.

So far we have discussed only dynamic models that enable us to determine whether an arbitrary input acceleration is physiologically tolerable, and for this work a linear model is perfectly satisfactory. In general, however, the natural frequency is also important in relation to free oscillation phenomena of the human body, such as those involved in determining the "take-off" velocity with which a man will leave the deck of a ship when the deck is subjected to an impulsive velocity change.

This last problem was studied by Frost Engineering for the David Taylor Model Basin and some preliminary results were reported by Payne(3), who found that the free oscillation frequency was different again from the two frequencies necessary to calculate P.I. and resonance phenomena. Thus the use of linear theory results in the derivation of three different frequencies, although we feel fairly confident that they all describe one, and only one mode of deformation. The reason for this, of course, is that the human body is essentially a non-linear dynamic system, so that the apparent linear frequency will vary with amplitude of deformation.

The non-linear theory associated with the three effects mentioned is examined by Payne(3), and it is found that a single non-linear equation will describe adequately all three phenomena, to within the accuracy of the existing data. Thus, on the grounds of brevity of statement alone, a non-linear representation of the human body is obviously to be preferred over the earlier linear models. Moreover, we can repose more confidence in the results of restraint system optimization when we use such a model, since it is evidently closer to the physical reality.

There are a number of applications immediately evident for this more refined model. For example it is current practice to calculate the trajectory of an escape system or capsule by assuming that the human occupant is a rigid mass. Subsequent rocket sled tests and aircraft firings with capsules containing anthropological dummies should (in theory) confirm these predictions because the dummy is also a relatively rigid mass. When the first firings are made with human occupants, however, the dynamic system is appreciably changed, in that the human occupants have their own degrees of freedom as dynamic systems.

In many cases this will be unimportant, but where the occupant's weight is a significant fraction of the total vehicle weight, and where there is a possibility of oscillations occurring near his fundamental mode of deformation, then the stability characteristics of the overall vehicle can be substantially modified. An example of where this could occur is during re-entry of a space vehicle, when any initial error in pitch or yaw can result in an oscillation of increasing frequency as the vehicle approaches the point of maximum $\underline{q}$. In those areas where the oscillation frequency coincides with a major natural frequency of the vehicle's occupant the amplitude of the oscillation could increase significantly above the calculated value.

So far in this historical survey we have confined our discussion to the human body. It has long been evident however, that this is only part of the problem and that the restraint system which connects the test subject to the source of acceleration is equally important. For example it was realized in the early days of ejection seats that the seat pan cushion was having a considerable influence upon the physiological effect of a given catapult charge and that an optimum cushion configuration and material should be established. Over a period of years the U. S. Navy experimented with dummies and human subjects on the ejection tower at ACEL and achieved significant improvements, and the USAF's Aeromedical Laboratory in a different program.

Early workers in the field pointed out that the determination of optimum cushion characteristics was a dynamic problem which should, in principle at least, be susceptible to theoretical analysis. Latham(8) conducted some simple analog computer studies along these lines and gave an indication of the benefits which could be obtained from a scientific optimization.

In general it is easy to show that the physiological effect of a given acceleration can be greatly modified by the dynamic characteristics of the system which restrains the test subject or vehicle occupant, and that the modification may be either favorable or unfavorable. For example, a poorly designed cushion in an ejection seat can increase the loads experienced by the occupant by a factor of two or three, and the same is true of a restraining harness if, for example, it permits the seat occupant to hang at some distance from the seat pan during ejection from a negative-g condition. On the other hand an optimum restraint system can attenuate a basically lethal acceleration to the point where it has no physiological effect.

It is obviously much more logical to determine the optimum dynamic characteristics by calculation rather than by experiment, since such a procedure is not only much less expensive and time consuming, but also avoids the inevitable experimental errors and scatter. In addition, it often happens that a particular type of acceleration-time history input, experienced at one corner of the operational envelope of the vehicle, is the most severe. In such a case the dynamic characteristics of the restraint system which are necessary to attenuate this particular input down to a tolerable level are different from the "design center" optimum which would be calculated for the nominal acceleration-time history imposed on the vehicle.

The calculations involved in determining the optimum restraint system are naturally rather laborious, particularly if a number of cases have to be considered. In order to eliminate this purely mechanical aspect, and also to provide a more precise substitute for the present practice of "eyeballing" acceleration traces, Frost Engineering designed and developed the restraint analog computer shown in Figure 10. The

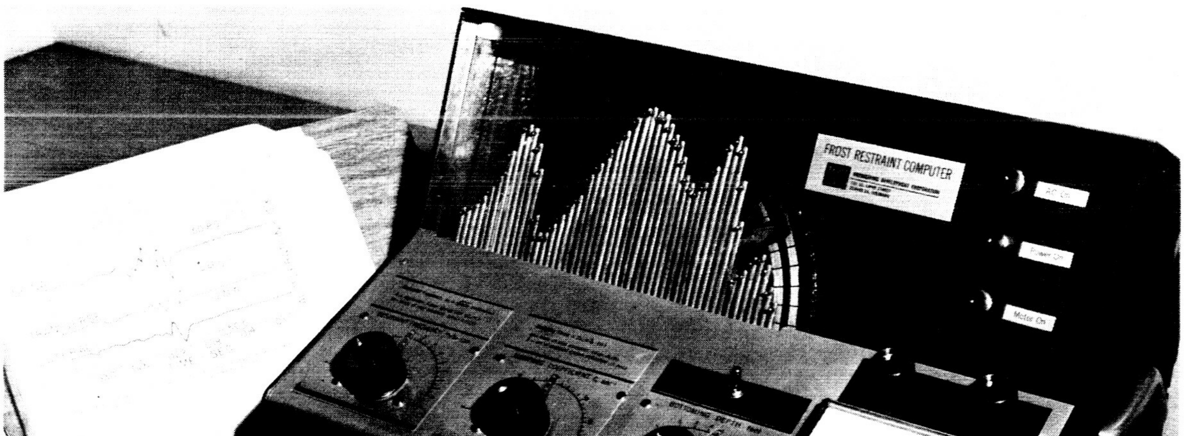


Figure 10. Prototype Frost Restraint Computer

array of rods on the front face of the computer can be moved vertically to plot a graph of acceleration input against time. The spacing of the input elements in time is such that any possible arbitrary acceleration-time history can be set up on the computer, and a meter gives the corresponding peak reading of the Physiological Index, which is a measure of the peak force which occurs in the human body.

In series with the dynamic models of the human body, which are presently limited to the transverse and spinal directions, the computer contains a linearized restraint system analog, the dynamic characteristics of which can be varied by the three large dials on the computer deck. These dials define the stiffness, damping and bottoming depth of the restraint system. If a given acceleration results in a P.I. so high that injury is likely to result, then the dynamic characteristics of the restraint system can be varied to find what combination will reduce the physiological reaction to the minimum value possible, compatible with engineering feasibility.

Prior to the introduction of this computer no satisfactory method existed for determining rapidly the tolerability of an arbitrary acceleration-time history. This restraint computer not only defines the tolerability of a given input in a meaningful way for the first time, but also enables it to be assessed in a fraction of the time required by alternative techniques. Moreover, optimum cushions and restraint elements can be defined in the design stage, rather than by expensive and time-consuming test programs.

A production version of this computer is now being designed by Frost Engineering, and is illustrated in Figures 11 and 12. Because of the wide variety of restraint systems in existence and the relatively fluid nature of restraint theory, it is inconvenient to attempt to package restraint circuits which are universally applicable to any restraint problem. For general work the existing circuits for a linear, bottoming system with

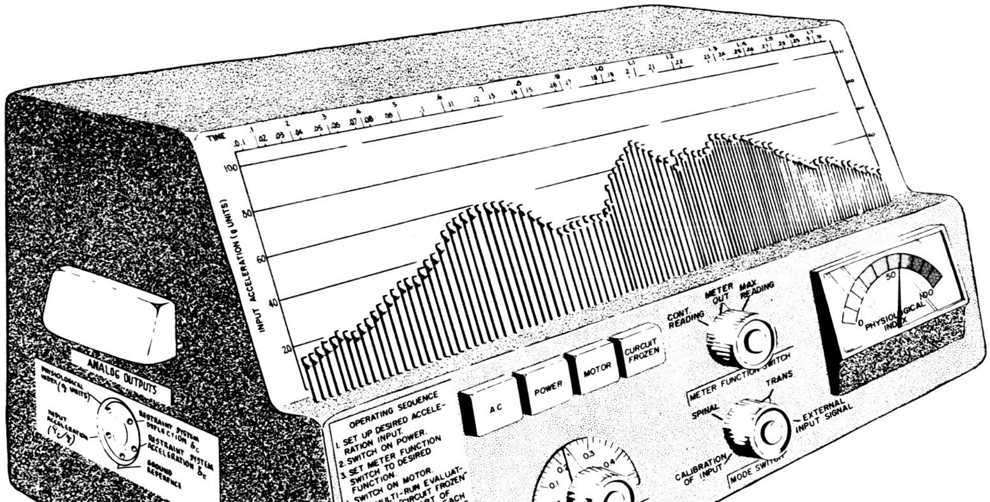


Figure 11. Production Desk Computer

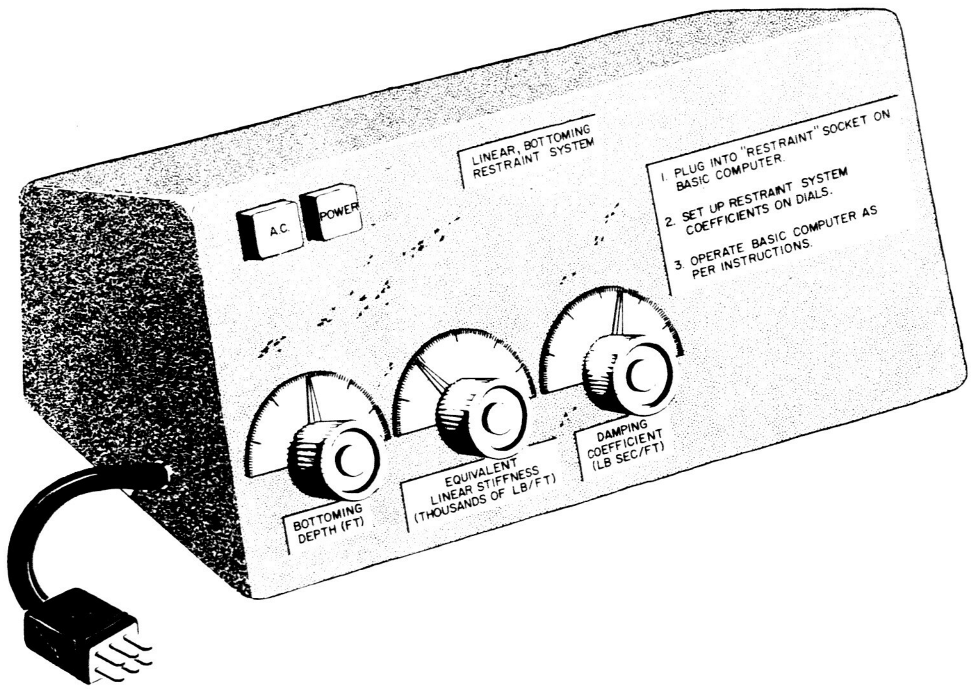


Figure 12. Linearized Restraint Module

one degree of freedom are probably the most useful but in special cases it may be desired to use one of the following:

a. Two separate linear systems, with different dynamic characteristics in the positive and negative directions.

b. Two systems as in "a" above, but with a substantial "dead zone" around the central position.

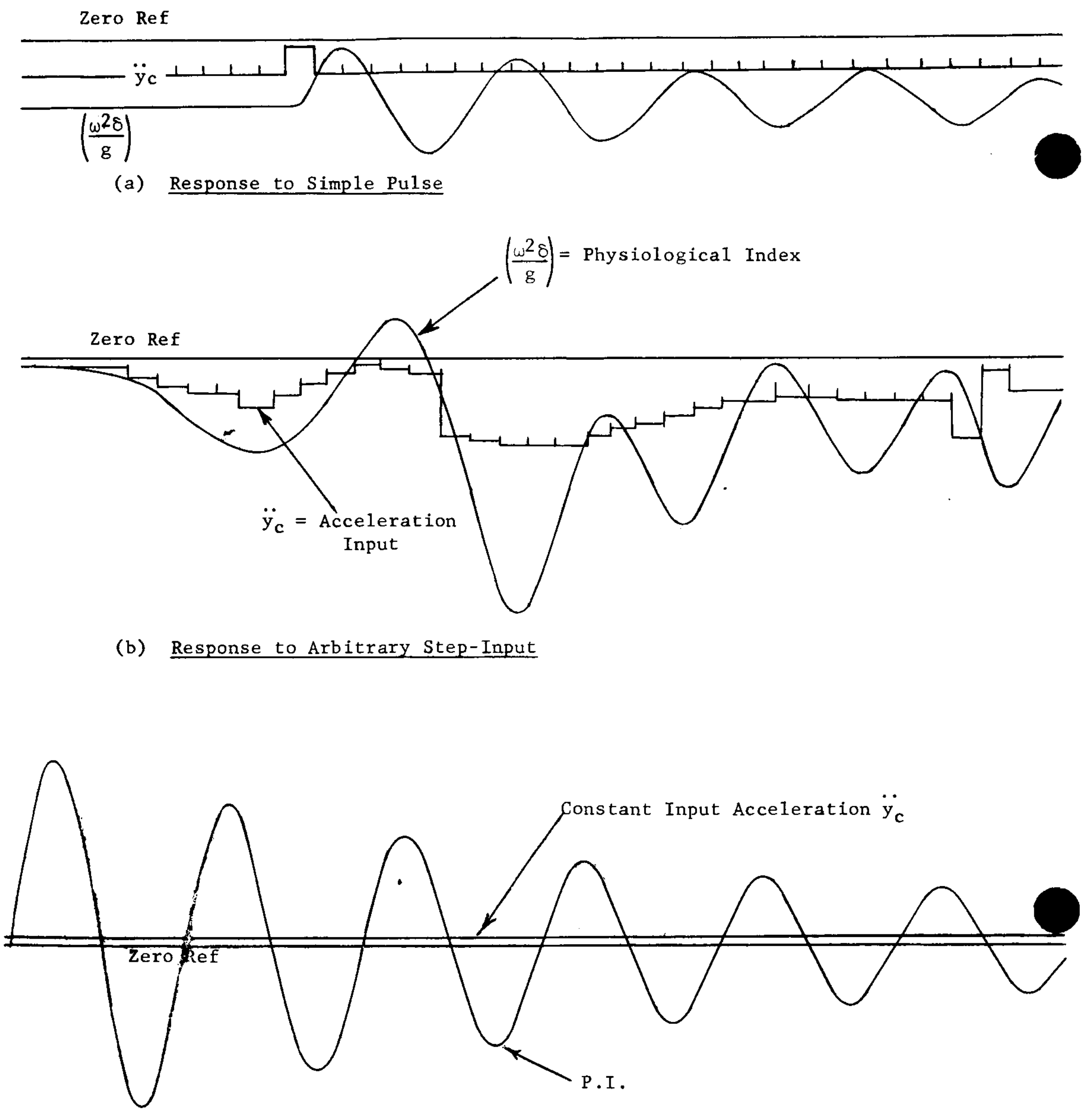
c. Truly non-linear systems for which the simple linear assumption is not adequate.

Since many users will require the circuit for only one of these alternatives, it was decided to manufacture separate "auxiliary computers" for the restraint systems and to confine the basic computer module to analogs of the human body, together with the necessary functions to perform the required analysis. Thus the basic module contains an 88-element input array for plotting an arbitrary acceleration-time history, dynamic models of the human body in the transverse and spinal modes, meter read-out of the P.I., a power supply, and provision for "fine grain" studies by means of a switch which reduces the time scale to 20 per cent of its nominal value.

This unit enables the physiological effect of an arbitrary acceleration-time history to be assessed for durations of less than one second and also supplies power for the auxiliary restraint computer which can be used to simulate the presence of a restraint system. For many evaluation problems only the basic computer is required of course, so that this division of function provides considerable savings to the user.

It should also be noted that the computer assembly is designed to be compatible with other computing equipment. Provision is made for direct read-out of all important variables on a recording galvanometer, and for integration of the computer into a larger hook-up, such as might be used in a study involving the overall dynamics of a vehicle. Worth mention also is that in both the basic computer and the restraint system modules, the passive circuitry is installed in the form of plug-in modules so that changes in the analog models can be made rapidly.

An example of the direct read-out capability is given in Figure 13, which shows continuous records of acceleration input $\ddot{y}_c$ and Physiological Index $(\omega^2\delta/g)$, as recorded on a Minneapolis-Honeywell Visicorder. Use of the basic computer as part of a larger hook-up is illustrated in Figure 14.



(c) Free Oscillation

Figure 13. Frost Restraint Computer - Typical Recorded Outputs

A number of aerospace companies have indicated interest in a somewhat more sophisticated restraint computer which will embody additional functions and will have the capability of handling, within the one package, several different generic types of restraint systems. The design of such a computer is currently under way at Frost Engineering, and the complete unit is illustrated in Figure 15. The input array on this model has twice the vertical resolution of the desk model, by virtue of sign changing switches under each element of the input array. It also embodies a scale change function that will enable the scale of the input acceleration to be changed by a factor of five, and has fully transistorized circuitry to eliminate warm up time.

In passing, it is interesting to note that the cost of any of these fixed circuit computers is substantially less than the cost to a company or agency of programming its own in-house computer, while the reliability is of course considerably greater. In addition experts in the field have pointed out that the method used for feeding in an arbitrary acceleration-time history on the Frost computer is much simpler and more rapid than can be achieved with any comparable device on the market at the present time; in fact, most commercially available arbitrary function generators cost almost as much as the entire Frost computer assembly.

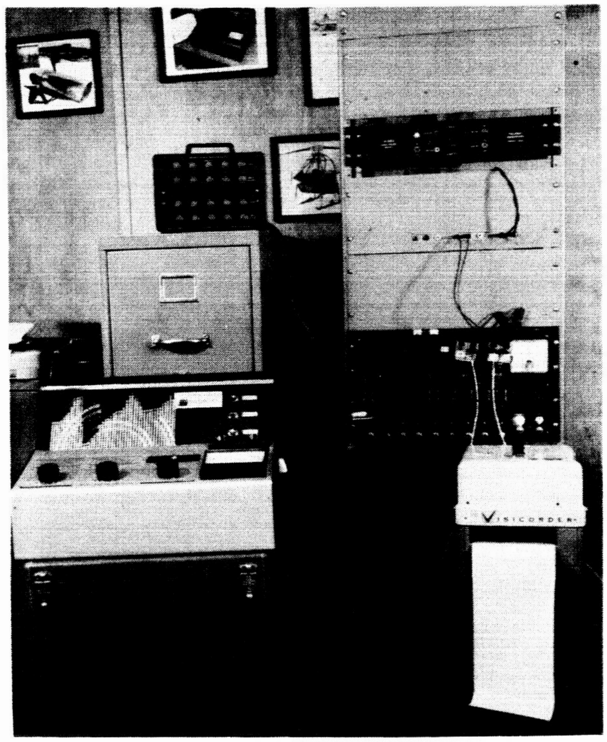


Figure 14. General View of Frost Analog Computer Equipment

The basic Frost Computer is compatible with general purpose analog equipment and can be used as part of a more complex problem hook-up, as shown in this illustration. The prototype Frost Restraint Computer is in the left foreground.

In reviewing the progress of this new technology, we see that in little more than two years we have advanced from very crude and often wholly erroneous "rules of thumb" to relatively sophisticated methods of estimating the physiological tolerability of any short period acceleration-time history. However, although this progress is quite spectacular in some areas, it is very patchy, and when we attempt to feed in numbers for specific cases we find that too often the information available is inadequate.

The reason for this unfortunate state of affairs is partly the newness of this technology and partly that much of the body dynamics research performed to date is the work of a few individuals who have not had the continuous support of a governmental agency. They have advanced the state of the art appreciably, but a great deal of work remains to be done to fill in the gaps in the present theory, and to obtain adequate experimental verification.

As a corollary that should be obvious, experimental programs carried out without any theoretical basis and subsequent validification are very often wasted. This is certainly so in physiological research work, and it is an unfortunate example of a field

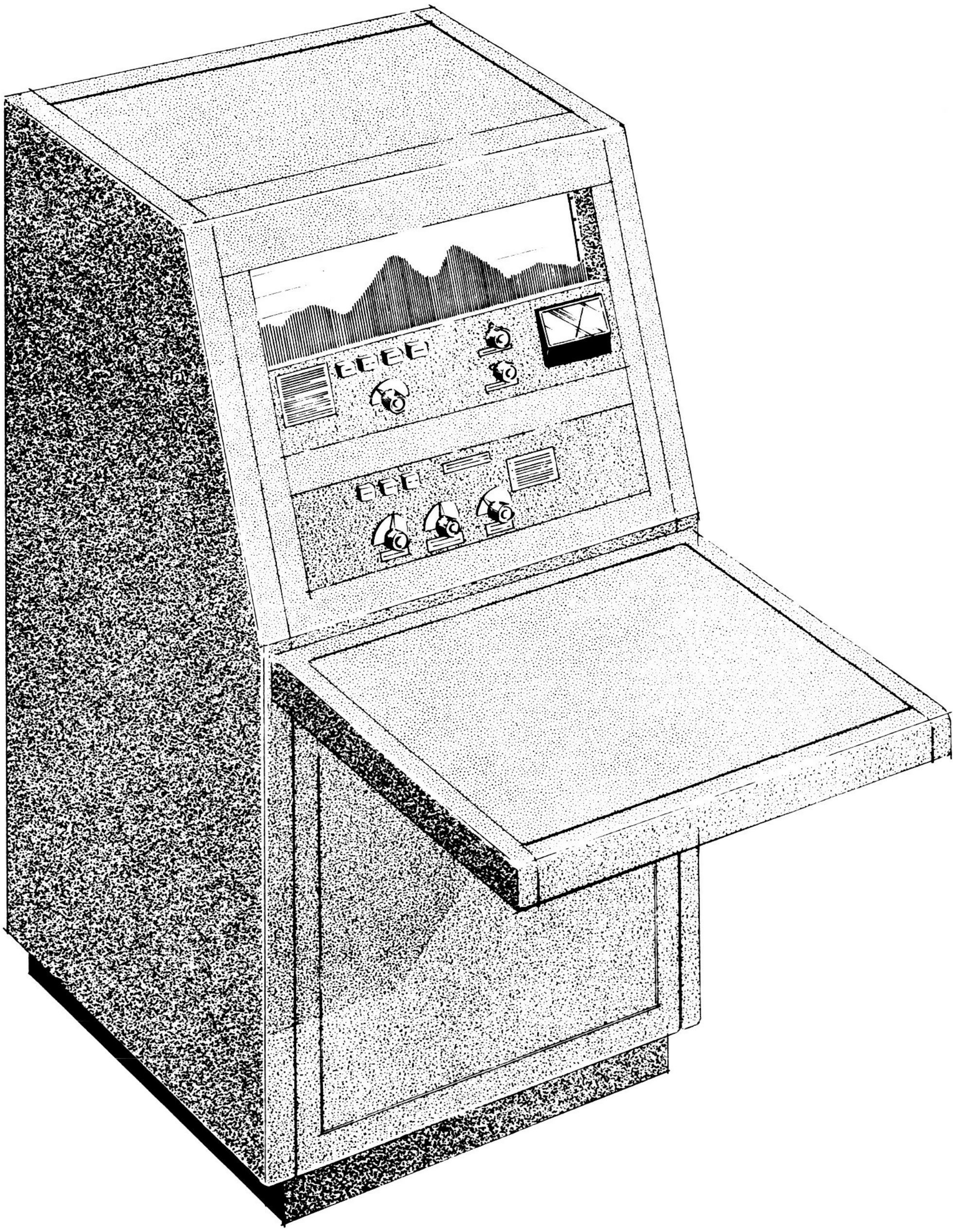


Figure 15. Console Restraint Computer

where expenditure of a great deal of time and money has produced relatively little as yet in the way of sound theory which can be applied in the design stages of a vehicle development.

Even when the theoretical and experimental gaps have been filled in to broaden the base of our knowledge at the present level, it is obvious that a great deal more work has yet to be done before the field of "body dynamics" will have become as precise and of the same general utility as most of the other technologies in science and engineering.

The Prototype Computer

The prototype restraint computer developed by Frost Engineering is shown again in Figures 16 and 17.

The decision to design and build this machine was not made until September 8, 1961, when the invitation to attend this symposium was received. Ten weeks is not a long period for a project of this type, despite our familiarity with the subject, principally because of the time necessary to design and manufacture some of the more complicated mechanical components. Thus the prototype is both relatively crude and somewhat larger than it need to be.

The front deck contains a meter which gives the Physiological Index ($\omega^2\delta_1/g$) in the positive sense only, and which is diode-protected for negative values. The models for transverse or spinal modes are switch-selected, and a second switch permits three modes of information read-out. In the "continuous reading" position the meter reads P.I. as a function of time, so that the output can be monitored continuously. In the "maximum reading" switch position the meter reads only the peak value, by means of a diode-capacitor peak follower circuit. Thus an accurate numerical value can be read off for the peak P.I. value achieved during a run, after the computer has completed its program.

The third and central position of the meter function switch shorts out the peak follower capacitor and disconnects the meter from the model circuits. This position is used when a permanent record of the output is required.

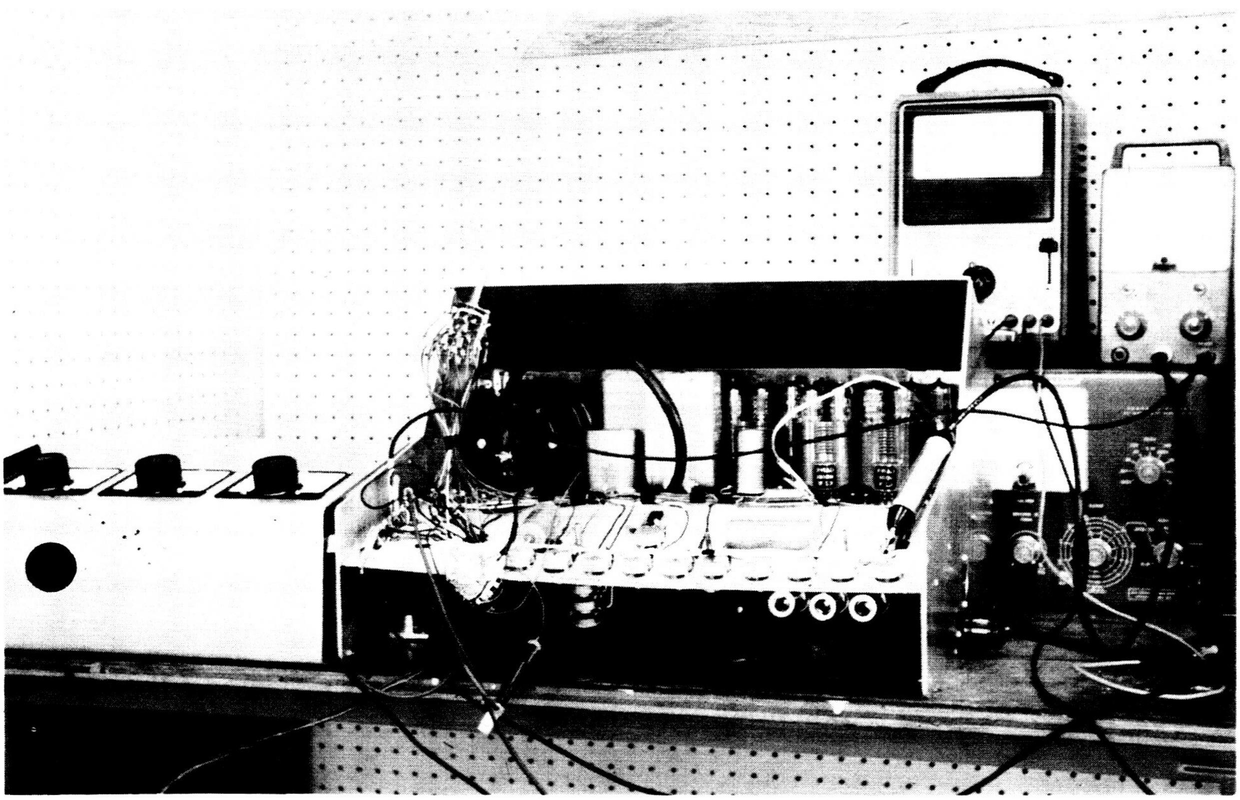


Figure 16. Initial Calibration and Adjustment of the Prototype Computer

Figure 17. Prototype Computer Being Used to Assess the Effect of an Experimentally Measured Acceleration Input

For this case, the signals for a recording galvanometer are obtained through a five-pin socket on the right hand side of the computer, these outputs being

a. Acceleration input $(\ddot{y}_c)$
b. Physiological Index $(\omega^2\delta_1/g)$
c. Restraint system deflection (δ_c)
d. Restraint system acceleration $(\ddot{\delta}_c)$
e. Ground reference

Thus three dependent variables can be continuously monitored together with the independent variable $\ddot{y}_c$, which constitutes the input to the computer.

It should be noted that by expressing P.I. in "g" units we can always check the calibration of P.I. against input, since for steady state conditions (all oscillations damped out) $G_c = \ddot{y}_c/g =$ P.I. Thus the d.c. reading on the meter should equal the d.c. input reading.

Restraint Circuits

The three large controls on the front deck vary the dynamic constants of the restraint system. Reading from left to right they control:

a. "Bottoming depth," which is the deflection necessary to make the restraint system become solid. This may be thought of as cushion thickness, when a cushion is the restraint element under consideration.

b. "Equivalent linear stiffness" of the restraint system.

c. "Equivalent linear damping coefficient."

The first two of these parameters are defined in Figure 18, while the damping coefficient is obtained from the known properties of the restraint system materials; a subject, incidentally, upon which a great deal more work needs to be done.

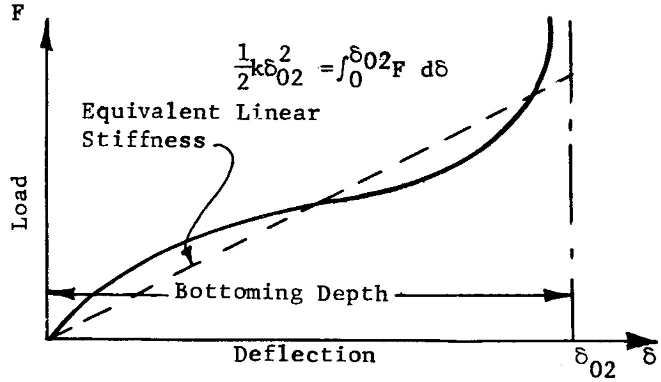


Figure 18. Definition of Equivalent Stiffness and Bottoming Depth

Input Circuit

The input to the computer is obtained with an array of linear potentiometers, the tops of which are transparent beads containing visual indication of their centers. These "cross-hairs" are aligned with the acceleration vs-time graph drawn or superimposed on the front panel, thereby setting the respective linear potentiometers to the values appropriate for the problem. Thus a real time continuous input is approximated by a series of "steps" which are read into the computer successively by a large rotary switch. The theory behind this approach is given in an appendix to this paper, together with a definition of the dynamic models used in the computer.

For convenience the rotary switch on the prototype is visible behind the transparent input panel, so that the status of the computation cycle can be seen directly.

The Model 1 Production Computer

The Model 1 computer illustrated in Figure 11 is designed for small quantity production, to sell at a price of $5,000.

Experience with the prototype has shown that the wide variety of restraint systems in existence demands a wider variety of analog circuits than can be housed conveniently in one package. Examples are:

a. the linear, bottoming system with one degree of freedom;

b. two separate linear systems, with different dynamic characteristics in the positive and negative directions;

c. two systems, as in "b," with a substantial "dead zone" around the central position;

d. truly non-linear systems for which the simple linear assumption is not adequate.

Since many uses will only require the circuit for one of these alternatives, it was decided to manufacture separate "auxiliary computers" for restraint systems, and confine the basic computer to the following seven functions:

1. 88-element acceleration input array.
2. Dynamic models of the human body with provision for variable damping.
3. Meter read-out of the Physiological Index.
4. Power supply for both basic and auxiliary computers.
5. "Freezing" functions.
6. Provision for external injection of the acceleration input signal.
7. Ultra-fast read-in, which reduces the time scale to 20 per cent of the "normal" value.

We envision that over the next decade our knowledge of the human body's dynamic response will improve considerably. Not only will future basic computer units handle impact and short period accelerations, but long period tolerance functions will be added, and possibly even mechanical impedance read-outs, as we improve our non-linear techniques.

The Model 1 Restraint System Auxiliary Computer

The Model 1 auxiliary restraint computer is illustrated in Figure 12.

This unit is plugged into the "Restraint" socket on the basic computer, and is then as fully integrated with it as the built-in restraint circuits on the prototype.

We anticipate that the single-degree-of-freedom unit illustrated will cost between $2,000 and $3,000 in small quantity production. More sophisticated restraint circuits will of course cost more, but most organizations will still find the cost negligible compared with designing their own circuits, and then developing them to gain the same degree of reliability.

Future Developments

We believe that the economics of human tolerance problems will generate a market for our specialized computers, and that at least 20 units will be needed, with a possible ceiling of 100 units. Therefore, rather than relying on government support for additional development, we hope that the sale of these computers will make the inevitable re[illegible]ment program self-supporting. Although it is obvious that no one is going to make any real profit from such a specialized and limited field, we feel that it should be self-supporting, if only because the cost per unit of making it so is negligible compared with the cost to each user of setting up his own program.

The immediate future will of course be concerned with putting the Model 1 units into production, so far as our work on computers is concerned. After this is accomplished we intend to develop circuits for long period acceleration tolerance problems, and more sophisticated restraint circuits.

One possible side-product of this work is a "real time" human model for physiological test work. This analog takes the signal received from accelerometers and in real time monitors the "Physiological Index," reading out the peak value on a meter as the test is completed. The advantages of such a procedure are obvious, and the saving in trace analysis time alone is very considerable. Frost Engineering has already developed a similar device for two of its customers, which permits the peak acceleration value experienced by an accelerometer to be read directly from a meter, and has already demonstrated that "instant data reduction" equipment can pay for itself in one series of tests.

With respect to the general field of body dynamics, a corporate objective of Frost Engineering is to maintain a continuing theoretical research program, and complete coverage of the relevant physiological research being carried out in this country and abroad. We hope to achieve this in four ways:

a. By seeking contractual support for fundamental research programs from NASA, USAF and USN agencies when the results of such programs, in report form, will be of wide general utility;

b. by company-funded research programs directly applicable to present and future computers;

c. by working as consultants with selected aerospace companies and research agencies directly concerned with the engineering problems associated with physiological limitations;

d. by closely following user applications of our computers, and participating in them when we can be of assistance.

In keeping with this stated policy and program, we solicit inquiries and requests for proposals from our co-workers in this vital and challenging field of research.

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APPENDIX

Accuracy of Step-Input Approximation

Fundamental Accuracy

It is shown in Reference 1 that when the base-line duration of an acceleration input is less than a certain critical time, then the precise shape of the input is unimportant; the velocity change (Δv) that it represents will be the only parameter affecting the behavior of the dynamic system upon which it is acting, and the equal area step function approximation shown in Figure A.1 can be substituted for the real input without loss of accuracy. For rectangular acceleration inputs the familiar tolerance limit graph of Figure A.2 gives an approximate definition of this critical acceleration duration.

$$\text{i.e.,} \quad \Delta t_1 = \frac{2}{\omega} \qquad \text{(A.1)}$$

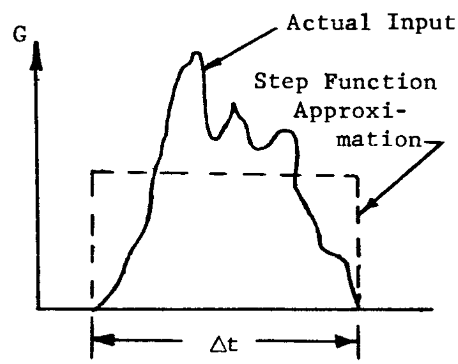

Figure A.1. Step Function Approximation to an Irregular Acceleration Input

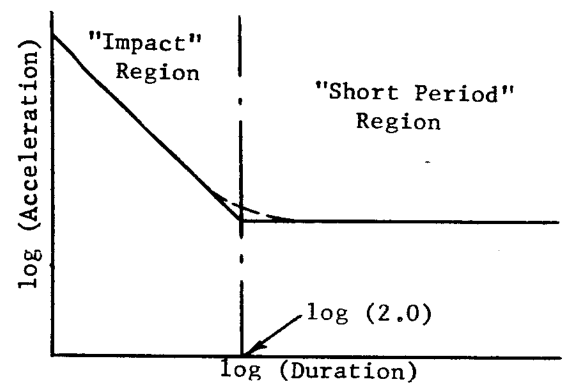

Figure A.2. Tolerance Graph for Rectangular Acceleration Input

The straight-line representation of Figure A.2 is however an approximation, the exact solution being represented by the dashed curve in the area of $\omega\Delta t = 2.0$. The exact equation is given, for $(\delta)_0 = (\delta)_0 = 0$ and for zero damping, by

$$\frac{\omega^2 \delta}{\ddot{y}_c} = \sqrt{2 - 2\cos\omega\Delta t} \qquad \text{(A.2)}$$

so that

$$G_{MAX} = \frac{\ddot{y}_{c\ MAX}}{g} = \frac{\omega^2\delta_{MAX}}{g} \Bigg/ \sqrt{2 - 2\cos\omega t} \qquad \text{(A.3)}$$

Now when $\omega\Delta t \ll 2.0$

$$G_{MAX} = \frac{\omega\,\delta_{MAX}}{g\Delta t} \qquad \text{(A.4)}$$

Thus we can define the maximum permissible duration for a step-function approximation by determining the value of Δt at which a specified error occurs between Equations A.3 and the simple velocity change criterior of Equation A.4. If $\underline{e}$ is the error, then

$$\frac{\omega^2 \delta}{\ddot{y}_c} = (1 + e)\ \omega\Delta t = \sqrt{2 - 2\cos\omega\Delta t}$$

$$\simeq \left[2 - 2\left\{1 - \frac{(\omega\Delta t)^2}{2} + \frac{(\omega\Delta t)^4}{24} - \frac{(\omega\Delta t)^6}{720} + \ldots\right\}\right]^{1/2}$$

$$\therefore \quad (1 + e)^2\ (\omega\Delta t)^2 = (\omega\Delta t)^2 \left\{1 - \frac{(\omega\Delta t)^2}{12} + \frac{(\omega\Delta t)^4}{360} - \ldots\right\}$$

$$(1 + e)^2 = \left\{1 - \frac{(\omega\Delta t)^2}{12} + \frac{(\omega\Delta t)^4}{360} - \ldots\right\} \qquad \text{(A.5)}$$

The numerical solution to this, for small errors, is as follows:

$\omega\Delta t$	= 0	0.5	1.0	1.5
$(1 + e)^2$	= 1.0	0.9794	0.9195	0.8266
e	= 0	-1.0%	-4.1%	-9.1%

The largest tolerable error is felt to be four per cent, for a simple pulse, even though overall accuracy will be considerably better, of course. Thus the width of a single "step" may be defined as

$$\Delta t = \frac{1}{\omega} \qquad \text{(A.6)}$$

where ω is the highest natural frequency in the dynamic system under consideration.

Considering total acceleration durations of up to one second, the total number of "steps" needed can now be defined by using the dynamic models defined in Reference 1. Here it is suggested that a finite damping coefficient ratio of $\bar{c} = 0.13$ should be used, rather than the zero damping assumption of earlier work, both for reasons of computer stability, and because we can anticipate that a lightly damped model will be nearer "the truth" than one with zero damping. As explained in Reference 1, we are currently unable to determine a "correct" damping ratio from test data because the data available is too limited.

The models of Reference 1 are illustrated in Figure 5a of section entitled A Survey of the Development of the Theory of "Body Dynamics."

The number of input steps required per second is equal to the highest natural frequency in radians/sec., from the criterion of Equation A.6. Thus we require

288 steps/second for the spinal mode
50 steps/second for the transverse mode.

Because of the practical difficulties associated with mounting 288 variable resistors on a desk-type computer, the Frost Model 1 computer carries only 88, enough for a total acceleration duration of

0.306 seconds in the spinal mode
1.76 seconds in the transverse mode.

Because of the practical difficulties associated with mounting 288 variable resistors on a desk-type computer, the Frost Model 1 computer carries only 88, enough for a total acceleration duration of

0.306 seconds in the spinal mode
1.76 seconds in the transverse mode.

Experience indicates that these durations are normally adequate, and of course, the short duration models are not supposed to be valid for durations in excess of one second. When longer spinal mode durations are required, the acceleration input can of course be read into the machine from an external arbitrary function generator, of which there are a number on the market. Alternatively, use can be made of the "freezing switch" which fixes all electrical voltages after the 88th step has been read out. The next 0.3 seconds of the acceleration-time history being assessed can then be set up on the input panel, and the machine restarted.

A newly developed optional feature is a time scale change, which speeds up the input read-out by a factor of five, giving

$$\left.\begin{aligned} \omega\Delta t &= 0.2 \\ \text{and } e &= -0.17\% \end{aligned}\right\} \qquad (A.7)$$

or less than five per cent of the error associated with the coarser input defined in Equation A.6. Part of the reason for this scale change capability is to simplify studies of very short period acceleration inputs such as occur in a number of Navy programs[3] but of course this feature can also be used in general work when extreme accuracy is required.

Errors due to Switching Transients

The wiper arms on the prototype Frost computer are spring-loaded in such a way that on leaving one contact surface they jump to the next, thus minimizing the time when the input voltage drops to zero. Transients as such do not have any effect of course, since the very nature of the analog circuit prevents it from responding to them. The only noticeable effect is that they reduce the total effective velocity change represented by the input, and most of this error can be calibrated out.

In the production computer, the use of rotary stepping switches, together with an RC low-pass filter, has reduced the effect of transients to negligible proportions.

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SOME CURRENT IMPACT STUDIES IN GREAT BRITAIN

J. C. Guignard
R. A. F. Institute of Aviation Medicine, Farnborough,
Hants., England

This paper gives a general outline of some of the work which has been done in recent years or is currently in progress in Great Britain in the field of impact acceleration stress. In making this presentation, the emphasis will necessarily be upon the work of the Royal Air Force Institute of Aviation Medicine, Farnborough, with which I am most familiar. In addition, I shall mention such other organizations as are known to be working in this field and with which it was possible to make contact during the short time available for the preparation of this report and bibliography. Neither the review of work nor the list of publications can claim to be exhaustive. Bearing in mind the sponsorship of this Symposium, it is perhaps important for me to make it clear at the outset that there is as yet no official British manned spaceflight program. There is, therefore, no work on impact or vibration being done in Government establishments which is specifically related to problems in space medicine. Acceleration research at I. A. M., for example, is directed solely to current problems in service aviation medicine.

Escape From Aircraft

Seat Ejection

There are still a number of outstanding problems in catering for successful escape from high-performance aircraft by means of the ejection seat, and these are the subject of continuing studies in Great Britain. First, we are concerned with the specification of acceptable force/time characteristics for upward ejection propulsion systems. The importance of the dynamic characteristics of the man-seat combination, and of the human body itself, in determining tolerance to severe, abrupt acceleration has long been appreciated by those concerned with the development of the British ejection seats. Practical experience combined with laboratory studies of human response to impact and vibration in the vertical axis (Latham, 1953, 1955, 1957b) led to the establishment of the current recommendation to the seat manufacturers. This is that the propulsion system should not apply a force exceeding 25 g, that the rate of rise of acceleration should not exceed 300 g/sec and, further, that a force of five g should never be exceeded in the first 0.01 second of ejection. Whilst theoretical objections can undoubtedly be raised to a directive in such a form, it has appeared to be a reasonably valid guide in practice. British operational experience with ejection seats now extends over some 12 years, and has recently been reviewed by Fryer (1961b). It is evident that the introduction of 80 ft/sec ejection guns, with a greater number and power of propellant cartridges, has increased the probability of the adopted physiological limitations being exceeded (Tolley, 1955) and has in recent years been associated with an increased incidence of spinal injury during ejection.

The Martin Baker Aircraft Company, Higher Denham (the leading—although not the sole—British manufacturers of ejection seats), are currently developing rocket propulsion for their seats, considerably to improve the chances of successful escape at low level and in unfavourable aircraft attitudes.* Using the rocket system, the force of ejection is considerably moderated, the plateau acceleration being of the order of 15g and the rate of rise not exceeding 200 g/sec.

Nevertheless, we still see an urgent need for more data on human response to forces applied in the spinal axis, of relevance not only to seat ejection but also to subsequent phases of the escape sequence. It is intended, therefore, to continue our studies of the dynamic response of the body to vibration and to carry out further instrumented shots using representative ejection systems on the static ejection test rig.

Survival Packs in Ejection Seats

An important factor in the application of the ejection load to the man (and one which has sometimes tended to be overlooked) is the nature of the mechanical linkage between the man and the seat. In most British ejection seats, the man sits on a non-rigid pack made of canvas with a top of leather or some similar material. The pack contains his survival equipment (solid), his rubber dinghy (evacuated and tightly folded, but nevertheless possessing some elasticity), and a thin upper layer of some compliant material to provide comfort. It was realized at an early stage of development that the pack acted as a spring and damper system interposed between the accelerating ejection seat and the man and that, if this pack was too flexible or rubbery, danger could arise both from poor ejection posture and from dynamic overshoot of acceleration acting along the spinal axis. Current British ejection seat packs are now made as stiff as is compatible with comfort in flight, and the upper layers are made of plastic foams with good damping characteristics. The top of the pack is shaped so that the buttocks are located to ensure a good ejection posture.

When a new operational pack is developed, it is the responsibility of the R. A. F. Institute of Aviation Medicine to test the prototype in order to determine its dynamic behaviour. The ultimate appraisal of a pack demands a live shot on an ejection test rig, but a convenient laboratory method has been developed to determine the damping characteristics of the complete pack by measuring the decay of oscillations excited in it by sudden loading (Glaister, 1959). This test enables an unsatisfactory pack to be rejected and redesigned before resorting to the ejection rig for appraisal: the risk of injury to human subjects testing packs on the rig is of course reduced by excluding from test packs which have been found to exhibit dangerous characteristics in the laboratory.

Ram Pressure

There is a serious lack of knowledge concerning human tolerance to ram pressure arising from the windblast encountered on ejection from high-speed aircraft. Whilst some data are available from studies in wind tunnels and on high-speed tracks using dummies, animals and human subjects—as well as from actual high-speed escape experience—all these approaches have practical limitations. A new method of studying

*Since this presentation was made, Sqn Ldr P. Howard, O. B. E. has successfully tested the Martin Baker rocket seat, ejecting from a Meteor aircraft at 250 ft and approximately 250 knots IAS on March 13, 1962.

the effects of ram pressure on man has been developed, in which human subjects are forced through water at relatively low speeds (up to about 32 ft/sec has so far been achieved) so that windblast is simulated by hydrodynamic drag (Fryer, 1961c). In Fryer's report of his recent experiments with this technique, he credits Capt. E. L. Beckman, USN (MC) with originating the idea of such simulation—an idea which arose out of the latter's experience in studies of the feasibility of underwater ejection from ditched aircraft.*

Using a specially constructed seat, propelled through water in the Rotating Beam Channel at the Admiralty Research Laboratory, Teddington, human subjects have been exposed to ram pressures of up to 7.2 lbs/in^2 (equivalent to 515 knots IAS in air). Injury levels have been attained in these experiments and various forms of head and limb restraint have been tested. Of particular interest is the development of a head cowl, attached to the seat, which reduces head buffeting at high speeds through the water. The method has proved instructive and promising, and proposals have been raised for more suitable apparatus for continuing these studies. The apparatus now in use has the disadvantages of poor acceleration (i.e., long periods must be spent achieving high speed) and inadequate peak velocity.

Aviation Crash Injury Research and Protection

The investigation of aircraft accidents in Great Britain is prompt and thorough and follows well-established procedures. Royal Air Force pathologists are normally represented on the official boards investigating major fatal accidents to civil as well as service aircraft. There is, accordingly, no lack of well-documented accident data. On the other hand, there is no experimental crash injury research involving human or animal subjects being carried out officially in Great Britain at the present time, although some live experiments have been done in previous years in connection with the development of aircrew crash restraint and escape systems (see below).

Seating and Restraint Systems for Aircraft

The high-speed track at Farnborough has for many years been used for impact experiments on human subjects. The majority of the tests carried out on it have been studies of forward crash restraint, involving decelerations of up to 16 g for 0.2 sec with rise-times of the order of 0.1 sec, and concerned with the development of safety harnesses for Royal Air Force and Royal Naval aircrew. It has occasionally been used also for investigations of the accelerations applied in carrier take-offs and landings (Ellis, 1955). In recent years, the track has been used for comparative tests of different types of aircrew safety harness. These include the "Z" type which is currently

*A series of experiments have been performed to determine the feasibility of using current Martin Baker ejection-seat systems to escape from ditched and sinking aircraft—a problem of special importance in Naval aviation. Whilst underwater ejection is admittedly a perilous procedure, it is marginally practicable with some systems (Beckman, 1959; Beckman et al, 1958, 1959; McNaughtan and Rawlins, 1960; Rawlins, 1956). With certain gun/cartridge combinations, however, the risk of severe injury is very high. A continuing program of research is therefore being conducted by Surgeon Cdr J. S. P. Rawlins and colleagues at the Admiralty Hydroballistics Research Establishment, Glen Fruin, in order particularly to study the effects of pressure waves generated on gun separation.

operational, and numerous variants of the combined parachute and safety harness. Live testing of their forward crash restraint is regarded as an essential part of the final appraisal of new types of harness intended for Service use. The accelerations applied in such tests (normally around 12 g for 0.2 sec) are moderate but realistic: we have not attempted to explore the extreme limits of human tolerance using elaborate whole-body restraint systems, but merely to appraise and suggest improvements to practical harness arrangements such as would be used in operational aircraft (Fryer, 1961d; Latham, 1957a).

The Mechanical Engineering Department of the Royal Aircraft Establishment, which operates the track at Farnborough, also carries out dynamic strength testing of aircrew harnesses and seats, using dummy subjects (for example, see Chisman, 1958). Safety harness fitted to aircrew seats (static or ejection) in R.A.F. aircraft are required, together with the seat and attachments, to withstand a deceleration of 25 g. The department is currently concerned also with the crash-worthiness of various types of seating used in passenger aircraft, again using dummy subjects in deceleration tests. The R.A.E. block dummy has usually been used in these experiments, although some experience has recently been gained with Alderson anthropomorphic dummies imported from the United States.

The Royal Air Force has pioneered in the use of backward-facing seating in passenger aircraft and the rearward-facing arrangement of passenger seating has been accepted policy in Transport Command for many years. The story of its introduction has been engagingly told by Dudgeon (1960). It is to be regretted that the promotion of the idea is meeting such protracted resistance in the commercial sphere.

Automobile Crash Injury Research and Protection

Apart from routine action by the police, road vehicle crash injury research on a systematic basis is conducted in Great Britain by the Road Research Laboratory, Harmondsworth (R.R.L.). Teams from the Laboratory make on-the-spot investigations of road accidents and it has recently begun a program of controlled impact experiments using actual road vehicles. A number of reports published by staff of R.R.L. are included in a recent bibliography of crash injury research compiled by Behr (1960).

Seating and Safety Harness in Motor Vehicles

The motorist's safety belt or, more correctly, safety harness is beginning to be regarded as a worthwhile item of equipment by the British motorist, although the campaign to promote its use is still far from approaching the success which has been achieved in establishing the almost universal use of the crash helmet by motorcyclists. Admittedly, the latter item began to be promoted some years earlier. It is to be hoped that, given time, the safety harness campaign will be equally successful. Unfortunately, much irrational opposition is still being voiced.

Within the past few years, a large number of firms—not all of the highest repute in this field—have begun to market car safety harnesses in Great Britain. At present, the safeguards against bad principles of design being embodied in harnesses sold to the public are unsatisfactory. The introduction by the British Standards Institution of the "Kite Mark" award (whose requirements include a static load test of the harness to 4,000 lbs) means that most harnesses now on sale to motorists are reliable from the strength-of-materials aspect. It is admitted, however, that static testing should

be supplemented by proper dynamic testing, in order to exclude faulty basic design in either the geometry of the harness itself or its tie-down. The British Standards Institution's Committee on Seat Belts* has inaugurated the establishment of a deceleration track upon which test vehicles will be used dynamically to test prototype safety harnesses, as well as for basic research. Speeds of 40 mph will be attainable on the 100 ft track at Hemel Hempstead, with stopping distances from three feet down to three inches (see Grime and Lister, 1961).

The British Standards Institution, the R.R.L., and other authorities concerned with road safety, are also studying secondary aspects of the design of car seating and safety harness. These include the durability of the buckles, straps and attachments of the harness, as well as the choice of webbing material for the straps. At present, some British harnesses are made of "Terylene" and others of "Nylon," the relative merits of which are still a matter of considerable debate.

Protective Headgear

A considerable amount of research has been devoted over many years to the development of protective helmets for Royal Naval and Royal Air Force personnel (Gabb, 1961; Rawlins, 1956b). Latterly, particular attention has been paid to the function and design of the different parts of the helmet, with special reference to the webbing harness and internal padding; and to the standardization of methods of testing protective helmets (Gabb, 1961).

The Road Research Laboratory has played a leading part in the development of crash helmets for motorcyclists (Chandler and Thompson, 1957) and have recently been working on a lightweight and discreetly designed anti-shock cap for motorists (Moore, 1961). Over the past few years, the R.R.L. has issued numerous research notes on these topics.

The War Office Clothing and Equipment Physiological Research Establishment at Farnborough is concerned with specialized forms of head protection; for example, brow-pads for sighting systems. Elwood (personal communication) and others have studied human voluntary tolerance to low-velocity impacts to the head.

Simulation of the Human Body in Impact Studies

Anthropomorphic dummies are widely used in practical testing of the various phases of operation of aircraft escape systems, as well as in the static and dynamic testing of parachute and safety harnesses. The R.A.E. dummy (Lovell, 1954) is that most commonly used, but, whilst it is approximately representative from the point of view of mass distribution, it does not in any way simulate the elastic characteristics of the human body. It is currently the standard dummy used in aerial trials of ejection seats and stabilization systems, although it is of interest that the Royal Aircraft Establishment has obtained practical advantages and useful results using reduced-scale mannikins for some purposes (Cobb, 1957, 1958).

For certain types of testing—for example, measuring the distribution of loads in parachute harnesses during simulated opening-shock—the R.A.E. dummy has been

*A number of interested authorities are represented on this Committee, including R.R.L., the Ministry of Transport, the Medical Research Council, I.A.M., and leading road safety and motoring organizations.

found clearly to be anatomically unrepresentative (see Jolly, 1958). British investigators have therefore shown interest in American dummies, such as the Alderson, which are anatomically realistic/and anthropometrically accurate. However, our objection to all the dummies so far available is that they are dynamically unrealistic because they do not embody the elastic and damping properties characteristic of the live human subject: properties which play a large part in determining the interaction between the man and his restraint or supporting system.

At I.A.M. we have recently undertaken to compare the dynamic behavior of various dummies in use with that of the human subject. The program has included instrumented shots on the ejection test rig and, also, resonance search tests on the vibrator, based on the hypothesis that if the dynamic response of a dummy were to resemble reasonably closely that of a man to steady-state vibration, then it would be valid to assume that the dummy was dynamically representative of the human body in impact testing. So far, the dummies tested have turned out to be quite unmanlike. Neither the R.A.E. prototype parachute test dummy (Guignard, 1961) nor an example of the Alderson type were found to vibrate with dominant mode at five cycles/sec, as does the human body: rather, they appeared to respond maximally at much higher frequencies. Recordings made during ejection rig shots of an Alderson dummy did not show the response typical of a live subject, but exhibited a poorly damped high-frequency ringing at about 25 c/s (unpublished I.A.M. data).

It has been suggested that it might be possible to produce dummies that were dynamically representative by incorporating within them tuned resonant masses so that, on resonance and impact testing, their behavior resembled more closely that of the human body. A theoretical study of this possibility is being made at the R.A.E. It is debatable whether it is worthwhile to attempt to create an all-purpose dummy that is, both anthropometrically and dynamically, an exact representation of a man; such a development must be prohibitively difficult and costly. Current British opinion tends to support the idea of making different dummies for specialized purposes. Such dummies might not look very human but would have the relevant parts of their anatomy correctly fashioned and present the correct impedance to dynamic excitation along a particular axis.

Vibration

(It was not originally intended to deal specifically with vibration research in this presentation but, in view of its general relevance to impact acceleration stress, and interest expressed during the Symposium, this section has been added in the preparation of the present text for publication in the Proceedings.)

British experiments on human responses to whole-body vibration at infrasonic frequencies began again (after a lapse of some years following Edwards' [1950] work) in the course of Latham's (1957b) analysis of human body oscillations during seat ejection. At about the same time, I.A.M. was also asked specifically to investigate the effects on man of sustained transient and harmonic vibrations in the infrasonic region (roughly 1 to 20 cycles/sec) in connection with projected developments in military and civil aviation. The human body-resonance studies begun by Latham were therefore taken further, using a standardized method of conducting resonance search tests on man, and proceeding to investigations of the influence of various intrinsic and extrinsic physical factors upon the human response to steady-state vibration (Guignard and Irving, 1960). Arising from these physical experiments, various physiological reactions to intense whole-body vibration of greater or lesser specificity have been incidentally

studied; for example, the effect of low-frequency vibration on the electromyographic activity of postural muscles (Guignard and Travers, 1959) and on respiration (Ernsting, 1961). Perhaps of more practical importance, British workers have recently conducted a number of studies of the frequency- and amplitude-dependence of performance decrement (mostly in terms of visual acuity) during whole-body sinusoidal vibration and similar modes of oscillation (Dennis, 1960;* Dennis & Elwood, 1958;* Drazin, 1959;** Drazin & Guignard, 1960; Guignard and Irving, 1960, 1961; Jones and Drazin, 1960).

We have a continuing interest in the physical aspects of human response to vibration and, in view of the dearth of actual experimental data (as opposed to theoretical construction) in this field, it is intended shortly to carry out a study of the body's linearity of dynamic response to steady-state vibration as a function of forcing acceleration.

At present, we have at Farnborough no facilities for applying other than sinusoidal vibrations to human subjects, but such a facility is in operation at the laboratories of Vickers Research Ltd., Ascot. This firm is using an electro-hydraulic positional servo system to reproduce complex or quasi-random waveforms from magnetic tape signals (Price et al, 1960). Within its limitations of bandwidth (0 to 20 cycles/sec) and amplitude (approximately 20 inches peak-to-peak is now usable) the machine can reproduce approximately the vertical accelerations recorded in flight through turbulence, with the addition of synthesized aircraft responses.

Conclusion

In aviation physiology, the most important field of research in impact acceleration stress remains, from the British viewpoint, the determination and specification of the forces which may safely be applied in seat ejection and other phases of escape from high-performance aircraft. With regard to the experimental work which still needs to be done, we would draw the attention of our colleagues working in this field to the urgent need for some agreement on and standardization of the parameters to be measured and the methods to be employed in the measurement and analysis of these forces. There are, for example, wide variations in methods of measuring rate of onset, or rise-time of acceleration in ejection tests.

In conclusion, I should like to thank the Space Science Board, and Dr. Hardy personally, for inviting me to make this presentation and for their hospitality. I have also to thank the Director-General of Medical Services, Royal Air Force, for permission to publish this review and bibliography.

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*Work of C.E.P.R.E., Farnborough.
**Work carried out at Wright Field.

Discussion

von Gierke: What is the preferred type of safety harness used in Britain or suggested for automobiles?

Guignard: On the whole it is felt by British authorities that the ideal type is the full harness with shoulder straps—provided that you have proper fixing-points on the vehicle. This is rather hypothetical, of course: it must be assumed, whichever type of harness is being considered, that the vehicle structure itself, the seating and the harness anchorages are properly designed. On the other hand, it is held by many of the organizations who are now fitting safety harness routinely in their cars—for example, some police authorities—that the lap belt with diagonal shoulder strap, similar to Dr. Aldman's Type 5, is a safe and acceptable compromise. Incidentally, I was most interested in Dr. Aldman's comments on women users: it is tending to be said in Great Britain that women prefer the full harness and are caused greater discomfort by diagonal straps. I don't know whether this reflects national differences in harness geometry or regional anatomy: I doubt whether anybody has yet made a proper survey on this matter (laughter).

The Road Research Laboratory have been interested and rather worried by reports emanating from the United States to the effect that lap-belts were quite adequate and that shoulder harness could be dangerous—particularly in producing whiplash injury following rear-end collisions. The latter danger, by the way, is one of the reasons for the disagreement among the manufacturers over the best type of webbing material for harnesses. We are certainly interested in American views on this subject.

To summarize, it is felt in Britain that the full harness is the ideal but that a properly fitted lap belt and shoulder diagonal is a very acceptable compromise. The police were among those who pioneered in the use of these, arguing that they often have to get out of their cars in a hurry in the line of duty, and that the full harness can be too troublesome to undo. The manufacturers also argue that harnesses will not sell to the public if there are too many straps to find and put together.

Question: Do you have any idea of the distribution of public use of the various types?

Guignard: I don't know the answer to that at the moment but could probably obtain an estimate for you. My own impression is that the lap-belt alone is rarely seen in Britain. The lap-belt with diagonal is now quite widely used, and you see quite a few full harnesses.

Aldman: It might be interesting to note that we have seen some accidents in Sweden involving the full harness with two shoulder straps, where one collar-bone was broken—after which you got completely free of the harness because you slide out of the other shoulder-strap and hit your head... We are revising our standards now and are going to approve the full harness type with two shoulder straps.

Guignard: Did this (the accidents mentioned) involve harness with a widely spaced four-point attachment or a V-type one with single shoulder strap attachment?

Aldman: A V-type.

Hardy: What about this so-called whiplash effect that there's been so much discussion about? Have you looked into it at all, Dr. Aldman?

Aldman: (Dr. Aldman indicated that the problem has been studied in Sweden.)

Haynes: I'd like to get something straight to our British colleague here. I am from the American automobile industry and, of course, we have a slightly different problem. Our cars are larger than the European cars and also we have to consider the use of the belt or the harness. Perhaps a shoulder harness is better than a lap-belt alone, but we have difficulty getting people to use even the lap-belt, let alone the shoulder harness combined. I don't think anyone will argue with the contention that there is some risk from whiplash effects if you do use the harness. Also note the fact that we (in the United States) have the larger car with more space, particularly for the passengers. We want to let (the occupant) swing in a defined position and perhaps provide proper structural protection in front of him, knowing where he is going to strike. We think perhaps this is as good a compromise as you can come up with right now; but we are not going to argue the point that shoulder harnesses aren't better (than lap-belt alone), particularly for the smaller compartments.

Guignard: In our cars, with their small compartments, the lap-belt alone would, we feel, often be lethal. On jack-knifing, there would rarely be enough room not to get a severe head injury.

Question: This also, I think, relates to the argument which favors long-stretching straps versus the short stretch. Even if you have more space, we would like to have as little stretch as possible.

Question: The usual type of hyper-extension or whiplash injury in this country is produced by one's car being struck from behind by another car: the problem is a little more complicated when there's a car in front of the car that is being struck. In other words, hyper-extension injury usually results from another car running into a car that is already stopped and, in this particular instance, your mechanism (i.e., harness) does not protect the patient at all. The head is pulled way back and might come forward, but the injury is mainly caused by the tremendous hyper-extension of the spine in the neck area, and, in some instances, this may be so severe as to cause an actual tearing of the spinal cord in the cervical area. Fortunately, this is a very uncommon finding; but I don't think your mechanisms here would protect against the so-called whiplash injury. They might give some protection in the situation where there is another car in front of the one being struck—that is, when the patient is hyper-extended and then hurled way forward. Then it might possibly help some. It might also be able to keep the body from being thrust forward... but the head and neck movements are not protected at all.

Guignard: My comment would be firstly that I am interested in your description of the mechanism of injury, because I have also heard it put forward that, following the rear-end collision, you get hyper-extension followed suddenly by rebound flexion—during which phase the injury is produced. I don't know what is your own view of that theory. I certainly agree that simple upper torso restraint wouldn't guarantee against whiplash injury. You would need head restraint, and I believe that some people in the American automobile industry are now fitting a built-up seat. I believe that it is generally said—although I cannot produce figures on this—that the rear-end collision is frequent in the United States but not often seen in Great Britain: at any rate, we do not see a lot of rear-end collisions of great severity—of the type occurring in your multiple turnpike pile-ups.

Hardy: I was very interested in the fact that your program on ejection seats is continuing at a high level of priority. Captain Smith, is there any parallel program of high priority in the Navy other than is going on at ACEL?

Smith: None that I know of...

Gell: I was also interested in this information. I had a long letter from Dr. Howard concerning these various problems, relative to proposals to the AGARD Acceleration Committee. It is just coincidental that many of these things you've talked about will be in our later panel as my contribution, in which we have made some suggestions for mutual work in the AGARD structure. I'd be happy if you would convey that information to Dr. Howard, and I will see if I can scrounge up a copy of this (statement of proposals) that you can give to him.

Guignard: Thank you.

Question: Some time ago I did a survey on the accident reports in the R.C.A.F.* on Martin Baker ejection seats, in which the spinal fractures were running at approximately 33 per cent. Although I was not able to obtain this same information from the British R.A.F. records, it was rumoured that the same ratio held in their case also. Are you doing anything to change the ejection concept to a rocket system, rather than a catapult alone, or is that work being carried on by Martin?

Guignard: The development of the rocket seat is being carried on by Martin Baker: the I.A.M. is in close liaison with the firm on the physiological aspects. I am sorry I cannot discuss details of this system at the moment.

Question: Regarding your point about the analysis of complex acceleration traces, I might mention that at Stanley Aviation we've had this similar problem; we have developed a dynamic model of the human body which tends to eliminate the uncertain way of looking at your acceleration trace by presenting a continuous input to our model and reading out a continuous output. In this way, we can in fact follow your trace, either by digital means or analog techniques, and therefore eliminate this dubious business of trying to estimate the rate of onset, peak acceleration level, etc. In fact we are, in a way, summing up the whole curve. Using the digital computer it is possible,

*Royal Canadian Air Force.

then, to look at the effect of ignoring certain peaks and (dips)* that appear in the trace. We have been able to obtain a smooth trace which gives us the same answer as if we had put in the real, complex trace.

Guignard: I've heard something about this work and we are interested in such an approach.

*Or 'oscillations'?

USAF IMPACT ACCELERATION PROGRAM AND FACILITIES

Major Willard R. Hawkins, USAF, MC
Lt. Colonel Rufus R. Hessberg, USAF, MC
Aerospace Systems Command

Introduction

Within the Air Force the biomedical, human-factor, or life-science studies are broad, all-inclusive, and definitely interrelated. On the ground, in conventional flying, and in space operations, accelerative and decelerative forces are ever present. This nation's greatest killer is crash injury, primarily due to automobile accidents. It has been said that one American is injured every 24 seconds, and one killed every 14 minutes, as a result of crash injury.

Under controlled situations we make accelerative forces work for us in most instances. In other controlled situations the stress is unavoidable and, though controlled, it is objectionable and man must be protected against it. The uncontrolled impact situation, with which this meeting is primarily concerned, is not only objectionable but, in spite of all safety factors, it is constantly plaguing us. To prevent injury we must know what produces injury, and thus much of our acceleration research is directed toward solving unknowns that have a direct or indirect bearing on the ultimate accomplishment of missions and objectives.

In aviation, as in space operations, the overall problem is the development of a system capable of supporting biological subjects and possessing size and weight characteristics which make it operationally practicable. Although they are not required in final operational systems, development stages will require means to measure the psychophysiological responses of biological components to the environment, and must provide methods of collecting the information for evaluation. Inherent in all aerospace flight is the exposure of vehicle occupants to mechanical forces. These forces must not be intolerable, and ideally will not limit performance of crew members. Limits of toleration to mechanical forces must be defined and established, and protective measures for aerospace occupants must be developed. With the equipment now on hand or under procurement these investigations will be continued and refinements initiated.

The purpose of our program is to establish criteria for design of manned aerospace vehicles in which accelerations, decelerations, buffeting, impact, and pressure differentials are to be encountered during any normal or emergency phase of flight. Present and proposed aerospace flight systems increase a thousand-fold the magnitudes, orientations, and combinations of the forces to which tomorrow's aerospace travelers may be subjected. Actual determination of human tolerance to each conceivable dynamic complex would be prohibitive in terms of time, cost, and effort. The alternative is the elucidation of sound principles of estimating tolerance to complex biodynamic stress. Such principles can be derived only from careful correlation of a small to moderate number of physiologic and psychologic responses to an equal number of

clearly defined biodynamic stresses. Development of a satisfactory method of estimating such human tolerance leads naturally to inescapable alternatives; development of methods of attenuating such stresses as must be applied, and developing methods of protecting the human or increasing his tolerance to those stresses that cannot be attenuated. An armamentarium of these principles and methods constitutes qualifications to advise the designers of advanced aerospace systems and insure compatibility of the performance characteristics of those systems with the physiological and psychological requirements of the human occupant.

Air Force Program

In general, accelerative research has preceded or kept pace with the requirements of the flight vehicle in which horizontal or conventional flight created physiological changes of short-duration g forces. Tolerances to the simple positive and negative g of conventional flight have been defined, and protective measures are in daily use.

The Air Force now requires definition of man's tolerance to the accelerative forces encountered during aerospace missions. The recent technologic advances in flight have outpaced our ability to acquire sufficient tolerance information—a situation partially attributable to the limitations of our present facilities for such research. The acquisition of acceleration-research data is used in establishing and/or limiting the engineering design wherein consideration of propulsion system, aerodynamic configuration, mission profile, and emergency escape influence the acceleration-force factors on the human occupant.

Such data have been obtained by the use of impact-accelerative devices such as the "Bopper", the "Daisy Decelerator" and the Holloman "High Speed Track." Among the data derived from past testing is the limit of 27g to forward-facing lap-belt-only deceleration. With the addition of the shoulder harness and an inverted V-belt for counter-tension, levels of 40g and upwards have been demonstrated to be tolerable. With the same restraint system the backward-facing subject has withstood 83g as measured with a chest accelerometer with no definite permanent injury. On this case the slope of the g-buildup was equivalent to 3,800g per second and the total duration of the g-time curve above a baseline was 0.04 seconds. Among the very few significant injuries have been one attributable to catapult cartridges which delivered more than their rated thrust, and one to preloading of the vertebral spine in compression by the shoulder straps where the subject's torso and neck-length exceeded those of the 95th-percentile man.

Investigations in meeting with human tolerance through force are carried out in three phases: (1) Using animal subjects, studies have been done at increasing force levels until enough data are gathered to establish satisfactory starting points for (2) human testing, which goes through a similar progression to the reversible injury level at which time testing reverts to (3) large primates used for determination of lethal levels. Frequently there are progressions within these progressions in regard to testing devices. The most frequent example of this is the progression from the Bopper to the Daisy Decelerator to the Holloman High-Speed Track Facility. The interrelationship of all tasks within this project is an obvious and established fact.

Tolerance to impact forces is being studied. The effects of short-duration, less than two-tenths seconds, force are being established in an effort to determine human-tolerance limits. To complete this area of investigation, rate of onset and organ

displacement are also being evaluated. Impact tolerance has been studied in the two common positions, forward-facing and backward-facing, both with the subjects strapped to an essentially conventional seat. Testing of prototype and production hardware has included the ejection-seat cushions, integrated harnesses designed for the B-52 and the F-104, the supersonic ejection seat for the F-106B, the supersonic escape capsule with the integrated harness for the B-58, the commercial passenger seat proposed by Belgium's Sabena Airlines, and a multitude of instrumentation and recording equipment for use on high-speed tracks. The ability to control headsnap voluntarily was studied under conditions of helmeting and deceleration commensurate with the Mercury escape-system g-time profile. Sixty vertical-drop experiments were made with human subjects in a simulated Mercury couch configuration (supine position) using impact velocities of 20 and 30 fps and shock-attenuating materials of aluminum and paper honeycomb respectively. Although a maximum of 64.8g over a time span of 0.08 seconds with an onset rate of 3,250g per second was recorded on a chest accelerometer, no adverse physiological reactions were noted.

The escape studies have centered on restraint, cushioning, and deceleration-attenuating devices using the Aeronautical Systems Division, Wright-Patterson AFB, Ohio's Inclined Test Facility, with speeds of 30 fps, peak-g loads of 38g and a rate of onset of 12,000g per second.

In evaluating human tolerance to the escape-force parameters, the effects of buffeting are being studied statistically so that a workable dynamic program can be instituted that will permit longitudinal-g application plus buffeting. Low-frequency random oscillations, low-frequency sinusoidal oscillations, and limited roll, as might be expected in ejections from high performance aircraft, are the areas of current interest. A prototype buffeting device is planned which in model form will prove design theories and be suitable for small biological-subject testing. Future full-size, man-carrying devices will be suitable for controlled laboratory experiments and sled testing. At Aeromedical Field Laboratory, Holloman AFB, New Mexico, interest is presently confined to those frequencies below 20cps, while higher frequencies are being evaluated at Aerospace Medical Laboratory, Wright-Patterson AFB, Ohio.

Much work has been done during the past 30 years in investigating the subjective response of humans to lower-level sinusoidal vibration, with wide ranges of results because the posture of the subjects, restraint belts on the subjects, and duration of vibration were neglected. Therefore, human tolerance to sinusoidal vibration from 1 to 15 cps for short-duration tolerance, one-minute tolerance, and three-minute tolerance, was established under controlled conditions. Physiological effects of sinusoidal vibration were also studied. Electrocardiographs were taken and no significant changes were elicited. Moderate increases in heart rate and blood pressure were observed. Blood studies revealed an apparent decrease in white blood count and a relative lymphocytosis following vibration at tolerance levels. Blood glucose, sodium, and potassium showed no appreciable change. In pulmonary ventilation studies there was up to 200 per cent increase in oxygen uptake and 5cps and tidal volumes were most affected at the thoraco-abdominal resonance range of 3 to 8.

The physical, physiological, and psychological responses of human subjects are now being studied in actual and simulated space-vehicle environments with the aim of establishing tolerance criteria for different types of random motions. Experiments will be designed to develop devices for increasing the tolerance limits to short-time duration, extremely high-intensity cycles, or transient accelerative loadings, present

during the firing of rocket stages, and of various low-level mechanical dynamic loads to be encountered during long-duration flights.

Tolerance to vertical vibrations (1 to 20 cps) transmitted to the semi-supine human subject was determined for a rigid chair and for a net seat (space couch). There was amplification of the vibrations which produced sensations referable to the abdomen and pelvic regions, in addition to the head sensations. As a result of these experiments, the net seat which could be used in space vehicles is being modified to damp the excessive resonances.

Further experiments were conducted to collect more data and confirm previous data relating to the existing short-time tolerance curve for human subjects in the sitting position. Individual variations were noticeable but, in general, conformed to predicted tolerance limits. Human tolerance to sinusoidal vibrations for five minutes in the 3-to-11 cps range is now being established. Preliminary indications are that amplitudes that determine five-minute tolerance levels are about 25 to 40 per cent of those that determine one-minute tolerance.

A continuing study of the mechanical characteristics of the human body is being accomplished, and the functional analog of the human body as a mechanical system has been refined. This model is very valuable in explaining man's response to vibration and impact forces and in arriving at protective means. Impedance measurements have been made on the human body undergoing sinusoidal vibrations in the semi-supine position on the rigid chair and on the net seat. It was shown that at subject tolerance levels the human body on the rigid chair responds similarly to a pure mass, while on a net seat mechanical impedance varied due to the presence of resonances of the man-seat system. The effect of this information resulted in the redesign of the net seat (space couch) to eliminate these resonances.

Studies are being conducted on the mechanical strength of tissues and relative displacements resulting from externally applied sinusoidal forces. The mechanical stress which material can withstand is determined by the effective force, frequency, and duration of exposure. Data obtained will allow the calculation of vibration and impact limits tolerable for body organs.

Continued research on the physiological responses of human subjects to vibration has included the recording of intra-arterial blood pressures taken at the radial artery and the evaluation of the sera and urine of human subjects for adrenal corticosteroids. Radial artery blood pressure under vibration shows the greatest change in the lower frequencies around 3 cps, apparently due to the reaction of the cardiovascular system to the mechanical resonance of the thoraco-abdominal system. Blood and urine samples were taken before, immediately after, and 24 hours after exposure to 3-minute-tolerance-levels at 1, 2, and 3 cps; 5, 6, and 7 cps; and 18, 19, and 20 cps. The response is a decrease in the adrenal corticosteroids after vibration with the greatest response per g of exposure and in the 5, 6, and 7 cps range. This is also in the range of lowest subjective tolerance.

Endocrinological and metabolic changes induced in animals by total body vibration are being studied. This research is aimed at detecting a quantitative indicator of stress from vibration by biochemical means and may provide a means of determining the early stress effects of vibration on the body. This biochemical indicator would also aid in judging the effectiveness of various protective devices.

Facilities

The description and characteristics of the existing devices are adequately covered in Motion Devices, which is Publication 903 of the National Academy of Sciences—National Research Council. Only devices that are new since the date of that publication, or modifications changing the characteristics, will be mentioned here.

New Device

The Aeromedical Field Laboratory at Holloman AFB, New Mexico, has acquired a Huge Shock Tester since publication of the NAS-NRC Report. The characteristics are the same as described on page 89 of that report for the Huge Shock Tester at Aeronautical Systems Division, Wright-Patterson AFB, Ohio.

Modification

The "Daisy Decelerator" has had 120 feet of additional guide rails installed which doubled the original length. A new water brake has been installed and the completed modification is presently undergoing acceptance testing.

The test parameters of the modified capability can provide the following:

Sled Velocity	-	5 to 175 ft/sec
Sled G	-	2 to 80 (100 with new sled)
Onset	-	100 to 8,000 G/sec^2 (Almost Square Wave)
Time	-	0.02 to 0.5 seconds

The following chart arranges the facilities by laboratory. Even though key personnel are listed, it is well to keep in mind that they can change, therefore the organizational component has been listed as a more stable reference. Only those devices under the control of the laboratories, upon which investigative activities are in progress and with which there are competent personnel associated, have been listed.

Proposed Devices

A new device is proposed for the Aerospace Medical Laboratory that will provide motion in five degrees of freedom by an electrically controlled hydraulic system. The purpose of the facility will be to explore human tolerance and performance under high-level angular and linear oscillations as they are anticipated during the re-entry phase of space vehicles, low altitude, high-speed flights of airplanes, and operation of escape systems at high speed. Simultaneous operation of all five degrees of motion with programmed acceleration patterns will be possible to simulate actual aerospace environments.

The device will have the capability of producing vertical linear motions from 0-30 cycles per second with a 9-inch double amplitude and a maximum velocity of 95 inches per second. The device will also be capable of producing linear, horizontal motions in the 0-30 cycles per second range with a maximum displacement of five inches double amplitude and a maximum velocity of 75 inches per second. The maximum load will be one thousand pounds.

Facilities

Device	Laboratories	Key Personnel
Vertical Accelerator	Aerospace Medical Laboratory Wright-Patterson AFB, Ohio	Col. A. Karstens
High Amplitude Vibration Machine	Bioacoustics Branch	Dr. H. E. von Gierke
Equilibrium Chair	Biodynamics Environment Section	Dr. R. Coermann
Shake Table for use on Centrifuge		
Vertical Deceleration Tower		
Dynamic Escape Simulator	Biophysics Branch Acceleration Section	Major A. Swan Dr. A. Hyde Capt. N. Clarke
Bopper	Aeromedical Field Laboratory	Lt. Col. H. Blackshear Mr. H. Feder
Daisy Decelerator	Biodyanmics Branch	Major E. Taylor Mr. R. Chandler
Huge Shock Tester	Holloman AFB, New Mexico	

The vertical angular motions of up to plus or minus 15 degrees pitch and roll and horizontal angular motions of plus or minus 30 degrees yaw will also be possible. Combinations of motions in all five degrees of freedom will be possible.

A Circular Track-Chamber

The laboratory testing of complex space flight missions today is not realistic because of the inadequacies of existing facilities. The multiple-stage boost at launch, for example, exceeds the capability of any conventional centrifuge as well as any long test track. To test the different space flight phases, such as launch, orbit and re-entry, and recovery, necessitates shifting the subject and hardware from one test facility to another.

The purpose of the proposed unit, planned by the Air Force Missile Development Center, Holloman Air Force Base, New Mexico, is to test the capability of man-machine systems. The anticipated test procedure encompasses the simultaneous and continuous testing of most of the parameters of the complete space flight history from launch through planet life to re-entry and recovery. It can be used to evaluate the calculated mission profile for human tolerance and it will provide the means for the biotechnological evaluation of man-machine units to insure that the equipment will support the subject when exposed to the actual space environment.

The Circular Track Chamber as proposed combines a 100-foot diameter circular track and 100-foot diameter hemispherical vacuum chamber into one composite test facility. The facility will handle payload factor products up to 2 million pounds. To cover this wide range two or three multi-purpose sleds are anticipated.

The support of the test unit by a sled offers sufficient flexibility to accommodate all equipment needed to apply the required test parameters such as oscillation, noise, etc., as well as telemetering systems. The maximum payload dimensions are 12 x 12 x 30 feet and the maximum payload weight (subject system) is 50,000 pounds. For maximum payload the load factor is 40g. For launch, a peak acceleration of 10g and an onset time of 140 seconds are possible. For re-entry (tangential) the maximum deceleration of 30g, a maximum payload of 50,000 pounds and an onset time of 50 seconds are possible. For 90 degree re-entry a maximum payload of 5,000 pounds is possible with 320g peak deceleration and a five second onset time. Water impact simulation in the chamber is such that 40g deceleration (peak) and 0.001 to 0.003 seconds onset time are possible.

The "Circular Track Chamber" described above represents one method of obtaining required parameters for future Air Force systems support. There are other approaches that have been suggested and studied. There is a recognized and urgent need for a device with multiple stress capability for biomedical research, but the dollar requirement demands careful and exacting study of the preferred approach and definition of performance to insure realistic correlation of the performance parameters of the device with future manned space systems.

Problem Areas

The major problem areas lie in manpower and funds. The problem of limited manpower available to support the Life Support area becomes progressively more critical for several reasons. The primary one is the need to provide increased technical support in the entire bioastronautics area to advanced systems study programs. Other reasons are the increasing scope of the program without a commensurate increase in authorized manpower.

The fund limitations for equipment, facilities, and devices have been discussed yearly and are mentioned here to be on record once again.

Major Change

The major change in the life support program has been the increased participation of bioastronautics laboratory people in the technical analysis and interpretation of advanced system planning documents generated as part of the Study Requirements and Planning Objectives programs. This has brought about a closer alignment of the overall life support program to future system requirements. In addition, the need for certain re-orientation and increased emphasis was detected. Almost without exception, our advanced manned system programs cite the protection (includes acceleration) and sustenance of man in the aerospace environment as major problem areas. The program, therefore, has been re-oriented to provide increased emphasis in these areas. Because of the need to operate within contractual fund limitations, the expansion in the protection and sustenance area will come about primarily by expansion of in-house research programs.

Future Plans

Future aerospace vehicles will expose crews to vibration and buffet levels endangering safety and mission capability. More effort is required to establish tolerance limits to random linear vibrations, angular oscillations, and long-time exposure. This work requires development of new motion simulators and new physiological and psychological test instrumentation. Launch, landing, and other maneuvers, including escape profiles of aerospace systems, expose the crews to high-impact and transient-acceleration loads. Additional effort is required (1) to measure impact loads of the various systems, including escape systems, and to predict them for future systems; (2) to establish human tolerance criteria; and (3) to derive protective principles.

Additional effort is required to establish tolerance criteria for combined environmental stresses, such as acceleration, weightlessness, buffet, vibration, etc. At present little is known on the effects of combined stresses. Space flight with its associated environmental and operational extremes pose the following problems: (1) tolerance to simple or pure acceleration, (2) tolerance to complex accelerations, (3) tolerance to acceleration concurrent with the imposition of other environmental stresses, and (4) tolerance to unique acceleration profiles of emergency escape.

The Air Force planning documents list the following efforts:

1. Studies will be extended to include human tolerance to vertical random motions as well as sinusoidal vibrations.

2. New principles will be investigated to reduce the relative displacements of inner organs and of muscular tissues with the skeleton.

3. Measurements of maximum strain on different types of tissue will be performed and calculations will be conducted to determine the traumatic limits of vibration and impact.

4. Physiological effects produced by defined patterns of acceleration utilizing the Vertical Deceleration Tower will be studied on animals and humans.

5. Work will be started on the physiological effects of low-level vibration with very long duration.

6. A new physiological test method will be studied to determine the objective magnitude of vibrational stresses. (Then methods have to be found to diminish these physiological effects and to increase the performance of the subject.)

7. The acute physiological effects of vibration will be studied to include the effects on the cardiovascular system (cardiac output, blood pressure, peripheral resistance, heart rate, and stroke volume), renal system (renal blood flow, glomerular filtration rate, filtration fraction and urinary volume), and intraluminal pressure changes (esophagus, stomach, rectum, and sigmoid).

8. Animal experiments are planned using radioisotopes to determine the effect of vibration on regional blood flow and blood brain barrier.

9. New principles of attenuating seating devices and couches for space vehicles will be studied to protect pilots against buffeting and impact.

10. Studies will be conducted to determine the optimum impedance curve for the man-seat system.

11. The effect on crew performance of relative vibrations between the pilot and the instrument display will be determined. (Then recommendations can be given as to how protective effects can be incorporated into a weapon system without impairing crew performance.)

12. Differentiate between the effects of onset ratio (the slope of the g buildup) and of absolute velocity change on impact tolerance. (Triangular onset shapes will be used and onset ratio will increase as absolute velocity change decreases.)

13. Determine whether change in velocity or in rate of application of g correlates with impact tolerance.

14. Study protection from deceleration injury by fluid immersion.

Recommendations

1. Continue studies to develop methods permitting physiological measurements during acceleration.

2. Complete work on the physiologic effects of forward acceleration, and develop prototype design of protective equipment.

3. Develop objective indices of acceleration tolerance.

4. Define tolerance to various acceleration vectors.

5. Define tolerances to spin and tumbling.

6. Catalogue acceleration and escape reference literature.

7. Study the effects of heat, vibration, and altitude on tolerance to acceleration.

Conclusion

Two resources have, to date, permitted testing in areas potentially too dangerous to justify using human volunteers. The large primate has proven an excellent subject for such testing by virtue of his anatomical and physiological resemblance to man. The anthropomorphic dummy has been the second recourse. Considerable amounts of research money and effort are being expended today in an attempt to make the dummy more comparable to man. The size, weight, and shape similarities are not sufficient in themselves for certain programs. The neck flexibility, density, skeletal strength, gas-filled cavities, and opacity to radiation are all under modification. Success in these additional facets of its iatrophysical evolution will necessitate subsequent internal instrumentation. Incorporation of functioning circulatory, respiratory, and nervous systems are an essential need. In short, the perfect dummy will never approach the live primate in anthropomorphism.

The alternative, and the biodynamicist is now at this crossroads, is to "go with the primate," and to concentrate research money and effort on his instrumentation. The end-product will be a test subject superior in many ways to the human. Internally

implanted strain gauges, radio-opaque organs, pressure transducers, and liquid and gas-filled cavities, thermistors, electrodes, accelerometers, and other physical and chemical sensors are conceivable. These input data can be telemetered to extra-corporal receivers, using power from internal batteries charged by induction, thus eliminating transdermal wiring. Performance, mood, and physiological alterations, structural stresses and strains, and organic and skeletal dislocations and injury can all be correlated.

The history of the impact-acceleration program shows a plethora of equipment testing limited by necessity to prototype and production hardware. Considerable information has been obtained concerning forward- and backward-facing impact tolerance and the practical aspects and procedures of impact testing. No testing in lateral, inverted, or intermediate positions has been accomplished because of priorities and equipment limitations; and no definitive investigation of the actual correlation of dynamic inputs, g-peak buildup, and duration to physiologic or pathologic results has been done. Determination as to which dynamic variant, force, absolute velocity change, or kinetic energy change is actually the limiting factor in impact tolerance, has similarly awaited time and means. These phases of the biodynamics program must be followed by incorporating the other variables such as vector orientation, subject orientation, harness configuration, elasticity and tension, ram air pressure, buffeting and oscillation. The infinite number of possible combinations prohibit reproducing each one in the test program, for already existing and anticipated combinations consume an inordinate amount of the available research effort.

Therefore, the program ahead must be limited to carefully defining fundamental principles of biodynamic interrelationships. These principles will, when verified, permit calculations of tolerance estimates when dynamic input data are provided. The accuracy of such estimates depends on the sagacity with which the program is pursued.

12572

PATH OF BODY TRAVEL*

Alfred L. Moseley
Department of Legal Medicine
Harvard Medical School

In studying automobile-accident deaths in a multi-disciplinary team setting, it is possible, in many cases, to predict the nature of occupant injury from the collision course and from the study of damage to the automotive structure caused by the impact of the occupants. Data of this nature, correlated with the individual medical reports and along with physical data of the accident, frequently allow a reconstruction of the kinematics of an accident, placing each occupant in his approximate pre-crash position. In certain medico-legal cases where there is a question as to who was actually driving the car, or where there is a question of whether the driver had slumped from the wheel, such a reconstruction is invaluable. This procedure is valuable in the single-occupant, single-car case in which there is need to understand the causes of death. When there are many occupants and multiple deaths, no survivor may have any useful memory. Objective procedures are of great value under these conditions.

After brief reference to rear-end and right-angle collisions, we will concentrate on the fixed-object collision.

Rear-End Collision

In the study of injury-producing accidents of all types, certain generalizations concerning the path of body travel have been described. The Abbott and Gay syndrome, for example, which has been described in and out of courts on many occasions, has come to be known as the whiplash injury. It occurs to people who are in a vehicle which is impacted from the rear. The long axis of the body up to the shoulder is restrained from rearward motion by the vehicle seat. The neck and head are not so restrained. The tendency is for the head and neck to remain essentially at rest, while the supported parts of the body are accelerated along with the vehicle. The acceleration of the unsupported head and neck lags in time the acceleration of the torso. This has the effect of a marked rearward motion of the head and neck with respect to the shoulders. Thus, the head and neck whip to the rear of the vehicle in the rear-end collision, producing the whiplash injury.

Right-Angle Collision

In the right-angle collision, the same general principles may hold; that is, the hit vehicle is moved in a direction conforming to that of the impacting vehicle. The occupants of the vehicle, which is hit, tend to maintain the spatial velocity, both in magnitude and direction, that they possessed just prior to impact. The effect is that

*From the Department of Legal Medicine, Harvard Medical School. Research carried on under Grant RG-6084, U. S. Public Health Service, Department of Health, Education and Welfare.

they are moving from their seated position to the point of impact on the side of the vehicle. Research in progress at UCLA by the team of Severy, Siegel, Gross, and Matthewson provides excellent information on the experimental right-angle collision.

We have observed some interesting phenomena in right-angle collisions. In several instances we have noted an imprint of the registration plate of the impacting car from two to five times along the side of the vehicle which is hit. Pattern injuries occur to occupants seated beside the doors under this condition. Multiple impressions of knob and handle projections are noted, giving rise to an interesting concept of the nature of the impact sequences. This concept is interesting in regard to what the insurance industry describes as the second accident. This description states that the first accident is the contact between one vehicle and another. The second accident is the contact between the occupant and the vehicle. This is the accident in which injuries are produced. Our data suggest that not just a second impact but many secondary impacts occur between the passengers and the vehicle. These multiple impacts give rise to many opportunities for the creation of initial injury and for making initial injuries more severe. This concept will be described in a later paper under the title, "Multiple Impacts."

Fixed-Object Collision

In the study of the fixed-object collision, we are dealing with a sudden deceleration in which some forward structure of the vehicle impacts an object such as a bridge abutment, a tree, a pole, a bridge pier, or some other similar unyielding object.

Variables Affecting Path of Body Travel

There seem to be four basic factors which influence the motion of the vehicle and the path of body travel during collision declerations. These are in addition to the more obvious factors of the human body weight and the velocity of the vehicle.

A. Parallel Path

Theoretically, the steady-state path of the body travel is parallel to the rectilinear motion of the center of gravity of the vehicle. Centrifugal forces associated with curvilinear motion also act on the occupants, but their effect can be neglected here. Thus, in a direct, central, fixed-object collision, the basic tendency of flight path would be straight ahead of seated position. Before, during, and after initial occupant movement, the vehicle exhibits complex motions so that the body-impact sites are the results of several factors operating simultaneously.

B. Vertical Motion

When the front of the vehicle impacts an unyielding structure, the rear end tends to rise up into the air. The weight transfer is due to the vehicle's center of gravity continuing its forward motion. Since the impact site is usually lower than the center of gravity of the vehicle, the front of the vehicle characteristically moves downward a short distance, while the rear moves up. The amount of angular shift in this weight transfer is related to the center of gravity of the empty vehicle, the deceleration of the vehicle, and the weight distribution of its contents. In some instances, the angular shift from the horizontal may be as much as 30 degrees. Some evidences of the angle of elevation can be seen in the rearward deviation of components such as the top part of the radiator. Impacts are observed in the hood structures near the leading edge of

the windshield. Some impacts are seen in the roof center adjacent to the trailing edge of the windshield.

C. Rotation in Horizontal Plane

In many fixed-object collisions, police officers at the scene are unaccountably alarmed because of the strange final position of the vehicle. There seems to be no sensible means by which to understand how the vehicle came to rest in its final position. In some instances, tracks are located, but tracks themselves do not lead to the position of the vehicle which leaves the officers in a quandary. If the impact-site is centrally located with respect to the lateral weight-distribution of the vehicle, then there will be no rotation. If the velocity is even moderate, yaw will occur with its center at the front of the vehicle. When the impact-site is off center, then the rear part of the vehicle will swing for some distance in a direction determined by the impact-site-center-of-gravity relationship. Under most circumstances, it is difficult to demonstrate more than a very small amount of rotation separately. Ordinarily, rotation and vertical motion take place simultaneously (Figure 1).

D. Deceleration Gradient

When the vehicle impacts a fixed object, the contact areas begin deforming and their forward motion is stopped. Adjacent structures on either side continue to move forward until the collision energy is dissipated. This differing rate at which the deceleration occurs is termed the deceleration gradient. The center of the area exhibiting the most deformation is the target point of the path of body travel. In experimental crashes with experimental dummies placed in the vehicle, it is interesting to note that considerable forward deformation may occur before the dummy begins to move. This means that vertical and rotational movement influence the attitude of the body in its flight path. If all forward motion of the vehicle has stopped when the body begins impacting vehicle structures, the contact velocity is the same for the body as it was for the vehicle against the unyielding object.

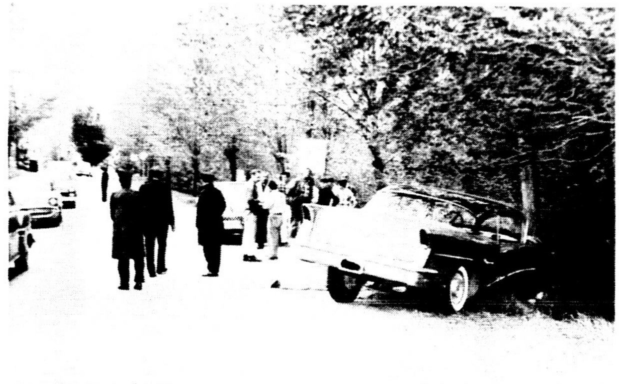

Figure 1. Vehicle shows a 24^{o} swing to the left after hitting a tree at 40 m.p.h.

E. Coriolis Force

In analyzing the kinematics of an off-center, fixed-object-type collision, it becomes apparent that immediately after structural contact and during the period that the body is in "free flight," the vehicle tends to rotate about the fixed object. This, in effect, causes a lateral displacement of the structure immediately in front of the oncoming body, in a direction away from the point of contact. The point of bodily contact is, therefore, apparently displaced toward the fixed object. We have found in collisions of this type that the bodily-impact point is normally found to be on a line drawn between pre-crash body position and the point of structural impact. It should be kept in mind that this apparently anomalous contact point is not caused by some inexplicable lateral displacement of the body, but by the rotation of the vehicle itself in combination with the other forces described. The actual flight path can be reconstructed

by utilization of relative acceleration techniques generally described in the concepts of coriolis acceleration and coriolis force.

Expected Path of Body Travel

As an operational matter, it may be said that the various combinations of forces involved in the path of travel of the unrestrained human body and the combination of several forces acting upon the vehicle in absorbing the energy at the moment of collision act so as to produce the following principle: If you know the seated position of a given occupant and the center of the impact site and draw a line in the plan-view connecting these two points, you can predict the longitudinal path the body will take in the deceleration. This is the expected path of body travel. Figure 2 shows a vehicle which has impacted a tree at the position of the right-front headlamp. Three occupants were seated in the vehicle: the driver, the right-front-seated passenger, who was sitting close to the right-hand door, and a rear-seat passenger who was sitting just to the right of the center of the rear seat. (For purposes of clarity, "left" in this and succeeding discussion will mean "left to the driver" and "right" will mean "right to the driver.") The Figure 3a shows the path of body travel of the right-front-seat occupant. He moved in a line directly toward the windshield. It was interesting to predict in this particular instance that the windshield structures had collapsed and had continued in motion when the vehicle impacted the tree, so that by the time the right-front-seat occupant reached the windshield area, there was no structure to impact. This judgment was confirmed by study of the body. The occupant received no frontal head injury. The rear-seat passenger moved in an angular direction, approximating the straight line from an estimated seated position toward the impact site (Figure 3b). Figure 4 shows

Figure 2. Impact site is at right-front headlamp. The collision course is the same as the point from which the photo was made.

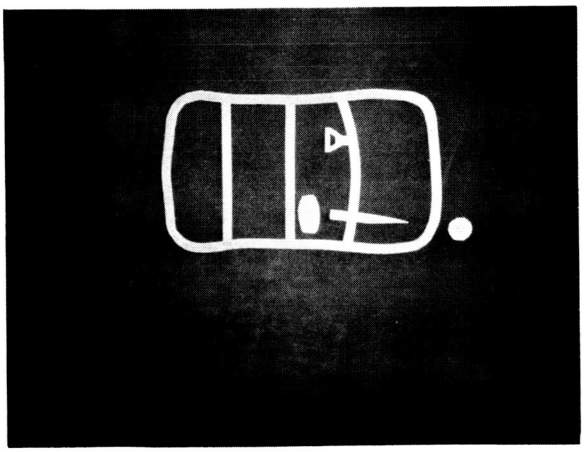

Figure 3a. Expected and observed path of body travel of right-front-seat passenger.

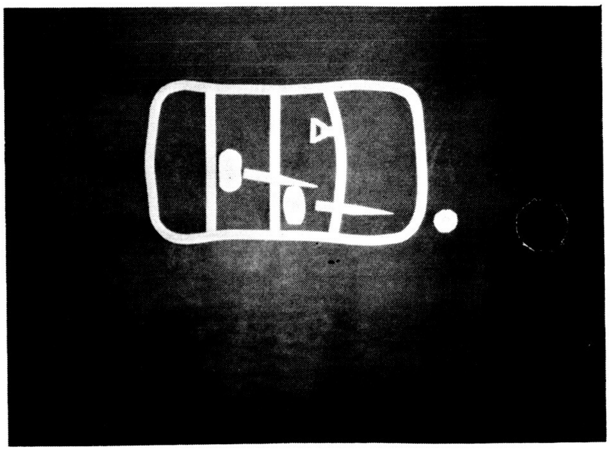

Figure 3b. Expected and observed path of body travel of rear seat passenger.

a picture of the opening to the trunk from the rear seat. The cardboard compartment-separation member might be presumed to serve the function of restraining any materials in the trunk. On this occasion, there was in the trunk an 18-pound, two-ounce wood box, containing a series of small tools. At the moment of impact, when the rear end of the vehicle moved vertically, it acted as a "launching platform" so that the box was lifted up in the air sufficiently to clear the vertical elevation in the floor which supports the bottom part of the rear seat. This box of tools came through the cardboard structure (Figure 3c), tore out the rear seat, impacted the rear-seat passenger in the left-rear chest, crushing his ribs, folding them inward, and producing multiple punctures of the lungs and heart. The passenger exsanguinated immediately. The box of tools continued toward the impact site and hit the right-front-seat passenger from the rear, inflicting a basal skull fracture and killing him instantly. The box continued on out through the open windshield area.

If you follow the same line of reasoning that has been illustrated for these two passengers, you will arrive at the conclusion that the driver in this circumstance should move at an angle conforming to the angular path of the previously described persons, should impact the steering wheel, the dashboard, and the windshield, in that order. This gives rise to the second general principle.

Observed Path of Body Travel

The accompanying diagram (Figure 3d) indicates that the path of body travel exhibited by the driver in this circumstance was not what would be predicted on the basis

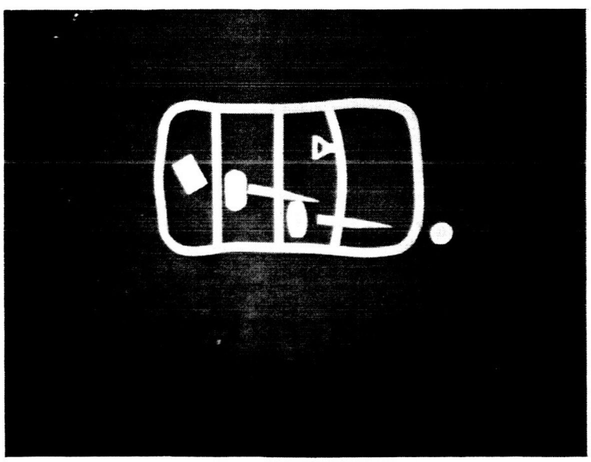

Figure 3c. Tool box in trunk came through the wall, impacting both passengers.

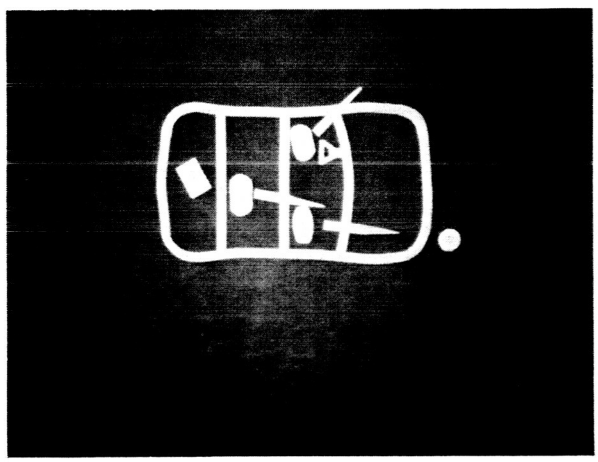

Figure 3d. Expected path of the driver is to the impact site. The observed path is not in accord with the prediction.

of seated position and impact-site. The special circumstances of this collision must be considered. The vehicle was travelling on a straight, flat section of highway and entered the area where there was a left turn. Three competent witnesses attest that the observed velocity was 39 to 40 m.p.h. At the curve, the operator of this vehicle did not turn to the left. Instead he went off on the right-hand side of the road where, for about 35 feet, there was a shoulder area not less than 12 feet wide. At the end of this area the road entered a filled area and the vehicle then began a slow turn while dropping down on an angle, riding along on the side of the fill. The driver of the

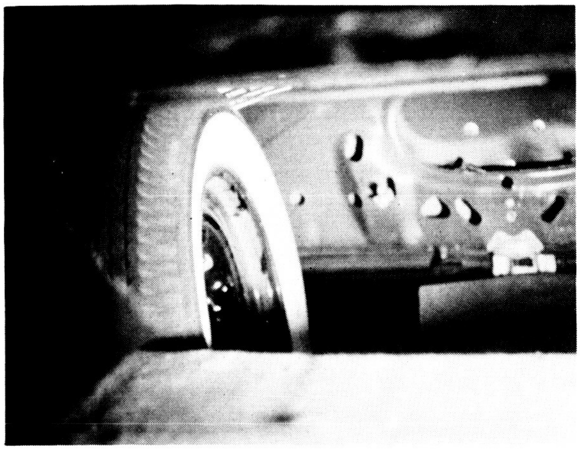

Figure 4. Opening made by tool box entering passenger compartment from storage area.

Figure 5. Two spokes of the steering wheel are printed on the driver's chest. On the low right chest is a pattern injury from a fractured horn ring.

vehicle made as sharp a turn to the left as he could, and succeeded in getting somewhat closer to the highway than he had been. If the tree had been a few feet further away, the impact would have been on the side instead of the front, and the deceleration forces would have been markedly reduced. At the time of the impact, the steering wheel was turned to the left. The right chest impacted the steering wheel to the left of the center post, so that the horn ring was imprinted under the right axilla. Figures 5 and 6 are photographs showing the chest of the driver and the steering wheel. You will note that the horn ring is fractured, and you can see the final position of the fracture of the horn ring with respect to the injury of the chest. This gives us concrete information concerning the degree of angular change in the steering wheel for the left turn, and it gives us the exact location of the body of the driver with respect to the steering wheel at the moment of impact. In addition to the deceleration geometry illustrated, other useful information can be derived from the evidence. Since the driver was in this position and making a left turn, we can make the objective judgment that he had probably not suffered a heart attack, and had not fallen into unconsciousness for some other reason. He was, in fact, conscious and doing everything humanly possible to avoid impacting the tree which he could see clearly ahead of him. This meant that the investigation could not be closed by a simple assumption concerning a loss of consciousness, but should rather be continued. What might have distracted the driver or taken the attention of all occupants so that the vehicle would leave the road where a turn should have been made? The finding was that there was an electrical fire under the hood of the car. The fire was producing acrid smoke which was coming into the passenger compartment. Our belief is that this smoke attracted the surprised attention of the driver and his occupants simultaneously. This, being a completely new experience to them, was distracting long enough for the vehicle to go off the road. There was a complicating physiological factor. The driver, himself, was not a smoker, but we found 10 per cent carboxyhemoglobin at autopsy. This was related to a hole in the exhaust pipe just forward of the muffler, and may have delayed comprehension time for a critical fraction of a second.

For purposes of orienting investigation on deceleration injuries, a discrepancy between the expected path of body travel and the observed path of body travel should alert the investigator that some usually complex factors are involved in collision

course. They require great precision and patience in studying the objective materials in order to arrive at a conclusion concerning the collision.

In another case, a vehicle northbound on an arterial circumferencial highway was out of control while in the high-speed lane next to a median, and travelled off to the right. The vehicle contacted the curbing and went over it for a distance. The driver steered so as to return to the paved surface of the highway. In the process of steering the vehicle back onto the roadway, he over-steered (Figure 7). The car went out of control and crossed three lanes to the left, crossed the median, and entered the southbound lane at an angle of approximately 80 degrees. At the moment of entering the southbound lane, another vehicle, moving in a southbound direction in the high-speed lane, impacted the right-front wheel and fender area (Figures 8, 9, and 10). The rear end of the impacting vehicle went up into the air and, while at a maximum point in this

Figure 6. The fractured horn ring which made the pattern mark in Figure 5. Mark shows the steering wheel was turned to the left at the moment of impact.

Figure 7. Vehicle went out of control to the right. Driver overcontrolled and crossed the median and was hit by a southbound vehicle.

Figure 8. Vehicle was hit at right angle by southbound vehicle after crossing the median out of control.

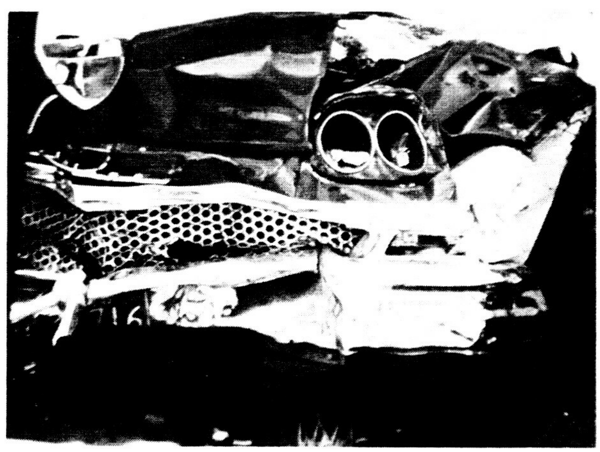

Figure 9. Front end of impacting vehicle. (See Figures 7, 8, 10, and 11.)

cycle, was impacted in the rear by another vehicle which went partially underneath it. Both of the occupants in the offending vehicle were ejected through the windshield area. It is interesting to see, as shown in Figure 11, that the third vehicle shows two evidences of head impact. The one to the left is an impact from an occupant of the vehicle. The laminated safety glass is impacted outward indicating the direction of motion. On the right is a second impact area. This impact is from the outside in. Hair is imbedded in the glass. It was of some scientific interest to make a determination as to which human had become the missile to impact the third vehicle. Hair was accumulated from the mortuaries and hospitals and subjected to microscopic analysis, which demonstrated that the head impact to the third vehicle was made by the head of the driver of the first vehicle. This path conforms very closely to what our prediction would be, based on the principles described above.

Figure 10. Steering wheel damage indicates path of body travel in hitting vehicle. (Figure 9)

Figure 11. Two head impacts in windshield. On the driver's right is a passenger impact. On the driver's left is impact made by head of driver of car in Figure 8. This driver was essentially a missile.

Summary

The above material has described the general characteristics of the directions of motion of the bodies of occupants inside vehicles involved in right-angle, rear-end, and fixed-object collisions. Human body weight and vehicle velocity are fundamental to the understanding of path of body travel. In addition to these, other general factors influence the motion of the vehicle and path of body travel during collision decelerations: (1) rectilinear motion of the center of gravity of the vehicle; (2) vertical motion of the vehicle; (3) rotation in the horizontal plane; (4) deceleration gradient; and (5) coriolis acceleration. The expected path of body travel in fixed-object collisions may be

operationally described as a path from the center of the seated position to the center of the impact site. The observed path of body travel of a given occupant may not conform to the expected path. In this event, some factors of a rather complex nature are involved in the collision course and require special investigation to identify them.

Conclusion

Sample clinical cases have been given to illustrate factors concerned with the path of body travel in the course of sudden decelerations. Three generalizations may be made:

1. The meaning of this and other clinical material is that the design of structures for the purpose of protecting an occupant during deceleration must take into account the location of the impact site. A steering wheel with a recessed-post design is very useful for absorbing some of the collision forces if the operator moves in such a direction as to impact it. If he does not move in this direction, the design has no bearing on his safety.

2. In view of the fact that multiple impacts occur between vehicles and structures which they hit, deceleration protection must take into account, not only an initial impact, but also the series of impacts which follow between the occupants and the vehicle.

3. Since deaths occur from missiles, a factor which should be of concern in deceleration geometry is the design of enclosures for storage or transportation of equipment. Any such materials may, under some conditions, become missiles at times of collision. Their paths would tend to converge at some point with the human occupant and cause serious injury.

Application: Advance Analysis

An evaluation of path of body travel may be carried out in advance of impact in any vehicle. The procedure would be to assume each possible impact-site in relation to the design and location of (a) compartment, (b) controls and displays, (c) occupant position, (d) storage area, and (e) protective equipment. Each of these would be considered in relation to the primary forces affecting the motion of the human body and the vehicle during deceleration.

12873

APPLICATION OF THE IMPACT SENSITIVITY METHOD TO ANIMATE STRUCTURES

Murray Kornhauser and Armand Gold
Missile and Space Vehicle Department
General Electric Company

Introduction

The method of presenting G-tolerance data in the form of a sensitivity curve was evolved at the United States Naval Ordnance Laboratory, White Oak, Maryland, in the mid-1940's[1]. Originally, this was a G-actuation presentation, since it described the performance of inertia-operated devices such as impact switches, which consisted, typically, of a mass on a spring which would be displaced by the acceleration pulse until it closed an electrical circuit. The G-actuation data consisted of the threshold levels of acceleration, at various durations, sufficient to actuate the switch. Also, the sensitivity-curve method of presentation proved useful when applied to structural failure, elastic and plastic deformation up to the failure point being analogous to the motion of the inertia mass restrained by its spring up to the point of actuation.

Application of this theory to man's impact tolerance[2] was attempted, but the lack of sufficient test points prevented firm verification of the theory. The mouse-impact studies described in this paper[3] were therefore performed on large numbers of animals under well-controlled conditions in order to evaluate the sensitivity-curve method of presenting G-tolerance data for mammals.

The Sensitivity Curve

A brief explanation of this powerful method of correlating and expressing dynamic performance is in order. Data are gathered by subjecting identical specimens to acceleration-time pulses of variable duration and amplitude. At each impact duration the minimum acceleration level that will cause damage is determined. Figure 1 illustrates the manner in which duration affects the peak acceleration necessary to induce a given amplitude-level of response. Note that the shorter the duration of loading, the greater the acceleration required for the same response.

Data are presented by plotting the parameters (ΔV and average acceleration) of each pulse which just succeeds in damaging a specimen. The cross-hatched areas (Figure 1) constitute the velocity change (ΔV), and average acceleration is equal to velocity change divided by duration. Figure 2 shows the sensitivity curve for a mass-spring system subjected to half-sine pulses. If the ΔV and average acceleration of an input pulse lies below or to the left of the curve, the specimen experiences no damage, damaging pulses falling above and to the right of the curve. Study of Figures 1 and 2 reveals the important features of the presentation: that the entire gamut of input pulse durations is represented on one curve. The steady accelerations and long-duration acceleration pulses appear on the vertical asymptote, whereas the short-duration pulses appear on the horizontal asymptote; the entire sensitivity curve representing the complete dynamic performance of the test specimens.

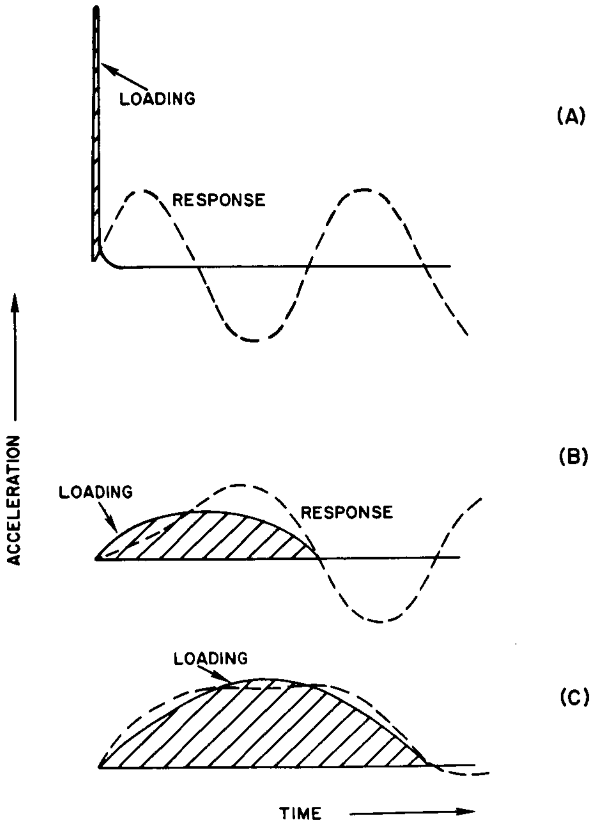

Figure 1

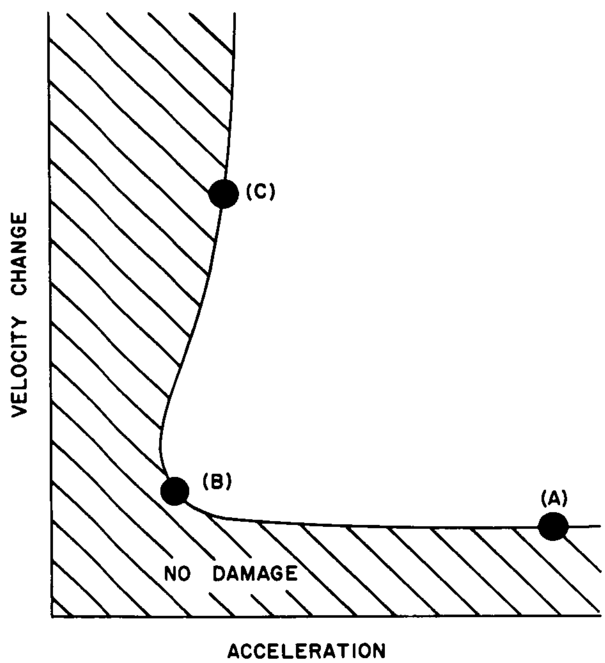

Figure 2

Velocity change and average acceleration are utilized as the significant parameters of the acceleration pulse to be plotted, rather than, say, rise time and peak acceleration. This is done because of the utility of the sensitivity curve which results, typically, when these parameters are plotted for either animate or inanimate structures. The utility lies in the existence of the two asymptotes, one of which implies that no damage occurs (regardless of velocity change or duration) unless a certain average G-level is exceeded, while the other implies that no damage occurs (regardless of G-level) until a certain velocity change is exceeded. The two statements taken together constitute a convenient way of expressing damage criteria: a critical value of velocity change and a critical value acceleration must both be exceeded in order to achieve a given level of damage.

Note that the sensitivity curve relates only to the parameters of the input function, no consideration apparently being given to response of the structure, coupling between mount and structure, and related parameters. This is done purposely, so that the sensitivity curve may be used in any application where the input acceleration-time history is known. The effects of coupling and structural response are already reflected in the fact that damage is produced, and there is no practical profit in learning the details of the response of each portion of the mount-structure system. As long as the identical mounting system for identical specimens is maintained, sensitivity curves should prove useful in predicting damage or no damage for any application which has known acceleration-time input characteristics.

Figures 3 and 4 indicate that the sensitivity curve for an identical set of test specimens is really a wide band in the long-duration region. Obviously, the location of the vertical (or long-duration) asymptote is a function of the shape of the input pulse. This implies the necessity of knowing more about the input pulse than ΔV and average acceleration. Note, however, that the horizontal (or short-duration) asymptote is

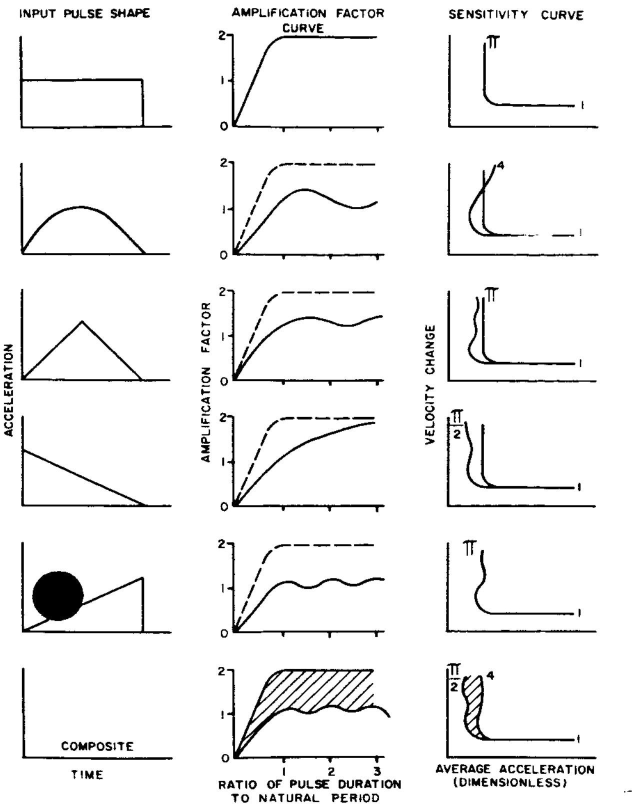

Figure 3

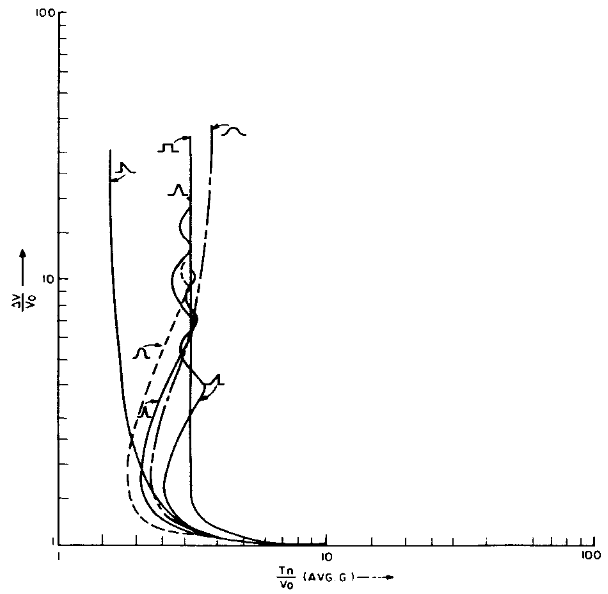

Figure 4. Theoretical sensitivity curves for mass-spring system (Avg. G).

independent of pulse shape, and may be specified by a unique value of velocity change. This constitutes a great simplification of some impact situations, the only specification being that the duration is short enough to be on the short-duration asymptote.

Figures 3 and 4 apply to mass-spring systems with one degree of freedom. For more complex structures, with several modes of failure, the results might appear as shown in Figure 5. In this case, the cross-hatched area would represent the "safe" zone of operation.

Application of the Theory to Man

Figures 6 and 7 present the experimental data on human impact strength (supine position) in the form of a sensitivity curve. The value of this form of presentation is evidenced by the two asymptotes, which enable one to state simply that the criteria of damage are 20-g and 80-fps velocity change, both of which must be exceeded concurrently for damage to occur to a well-supported human in the supine position. Of course, this statement is a gross oversimplification, and it should be refined whenever there is sufficient information available to warrant refinement.

Major sources of data are indicated in Figure 7 by the cross-hatched zones, which correspond fairly well to the "zones of impact" discussed above on a theoretical basis. Zone No. 1 of Figure 7 is the short-duration impact zone, the data for which are collected sporadically as a result of survival of air crashes and suicide attempts[4, 5].

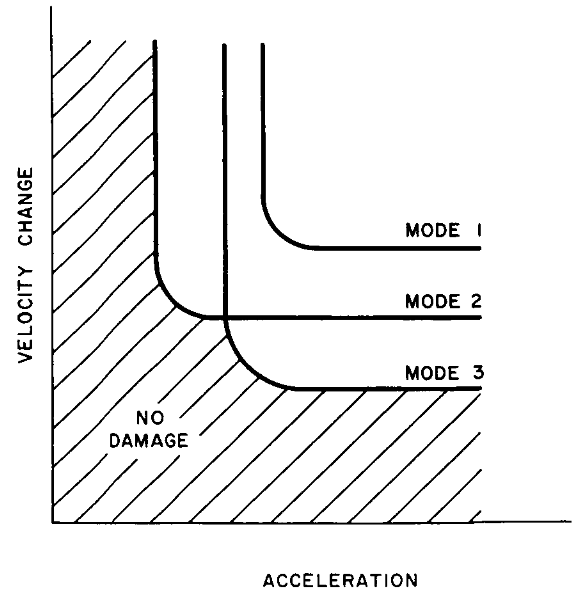

Figure 5

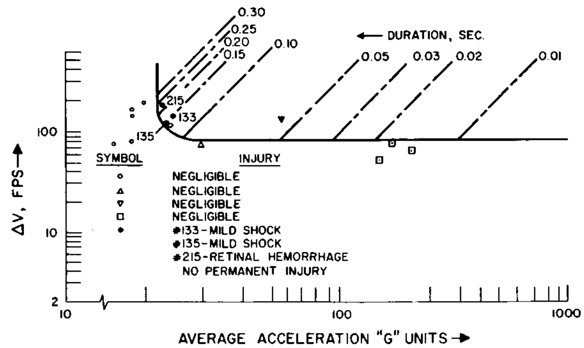

Figure 6. Human impact sensitivity curve.

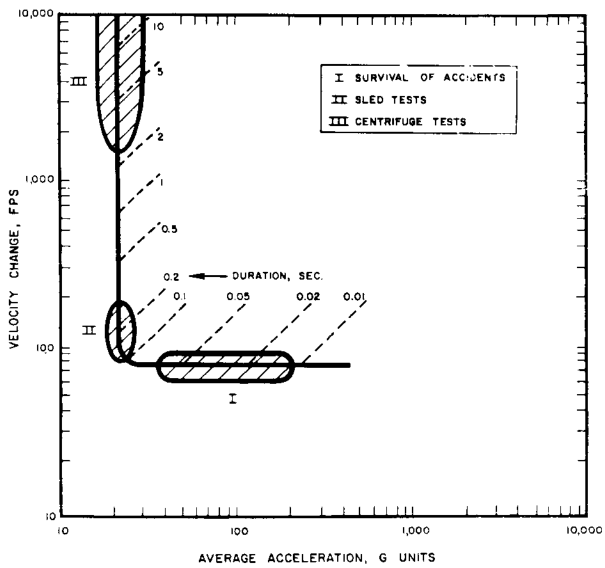

Figure 7. Human impact sensitivity curve.

As a result of this unsystematic collection of data, the short-duration asymptote for man is based on only five or six datum points. It was because of the uncertainty that the experimental points truly lie along the horizontal asymptote predicted by theory that an experimental mouse-impact program was instituted at the Missile and Space Vehicle Department of General Electric.

The Mouse Impact Tests[6]

Strain C-57 male black mice weighing between nine and 22 grams were used as test subjects. The mice were restrained in one-inch diameter clear plastic tubes, which were then loaded into a carriage and dropped freely from measured heights (Figures 8 and 9). The carriage was guided by a cable connected from the top of the drop-shaft to the floor. Molded lead stopping devices, designed to permit only single-phase acceleration pulses having slow to moderately rapid rates-of-onset, were positioned on the anvil. Oscillograms of these pulses were obtained as in Figure 10. Mice were oriented in the prone position so that deceleration forces were applied transversely to the spine (back-to-chest).

Average deceleration was calculated both from the oscillogram records, and from the velocity change (impact velocity plus bounce velocity) and stopping distance, the latter indelibly recorded in the form of the crushed lead slugs.

In preliminary experiments, mice demonstrated an all-or-none response to short-duration deceleration, either expiring within 30-40 seconds or surviving indefinitely without observable permanent injury. Ability to survive impact was therefore selected as the index of tolerance.

The lower curve in Figure 11 was constructed from plots of 11 groups which demonstrated a high incidence of survival (91-100 per cent by direct observation). Group sizes ranged from 12 to 23 mice. Asymptotes were determined from the curve at approximately 27 feet/second (velocity change) and 650 G (average deceleration). Asymptotic values of the upper curve, constructed from five points representing groups in which a low rate of survival (0-8 per cent) was observed, were approximately 45 feet/second and 1,970 G. A total of 329 mice were dropped in the two series.

Figure 12 shows the responses of 25 groups, ranging from 12 to 20 in number and totaling 427 animals, which experienced intermediate incidences of survival. Percentage of survival in each group is indicated. Those groups having high rates of survival generally occur near the lower curve; conversely, those with low rates of survival appear near the upper curve. This information was useful in the design of experiments which ultimately provided data for the two tolerance curves.

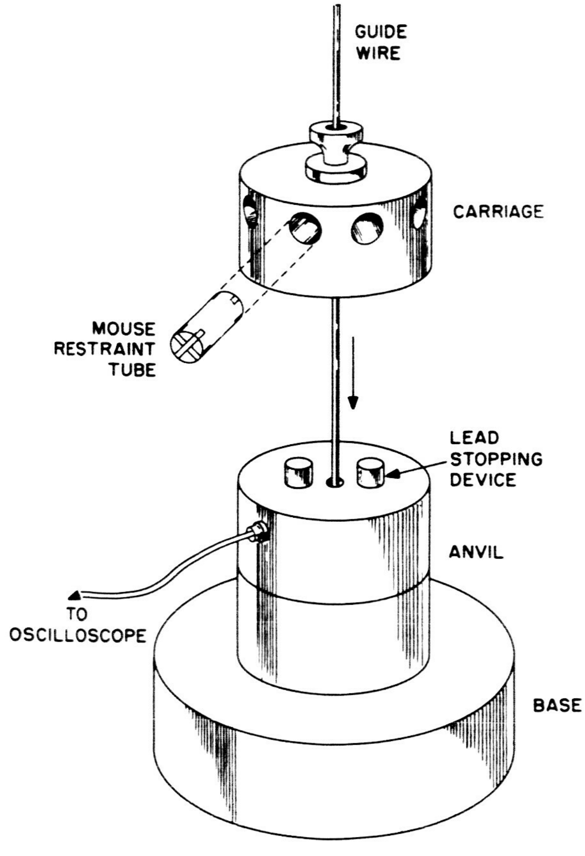

Figure 8

Figure 9. Test container and mice.

Brief periods of generalized convulsions were often observable in animals which failed to succomb instantaneously. Bloody fluid in the oronasal region and marked extension of the scrotum were also occasionally seen.

Gross and microscopic post-mortem examinations were conducted on 25 dropped mice selected at random from various groups. Of these, 11 died within several seconds after impact; the 14 survivors were sacrificed with ether. Hemorrhagic lesions were commonly detected in the lungs (Figure 13) and to a lesser extent in the spleen and on the surface of the cerebrum. Lesions were more severe in the high-mortality series. In only one mouse, which succumbed on impact, were hemorrhages noted within the brain substance (Figure 14). There were no abnormal findings in the heart, liver, and kidneys. Organs and tissues of three undropped mice sacrificed with ether were normal.

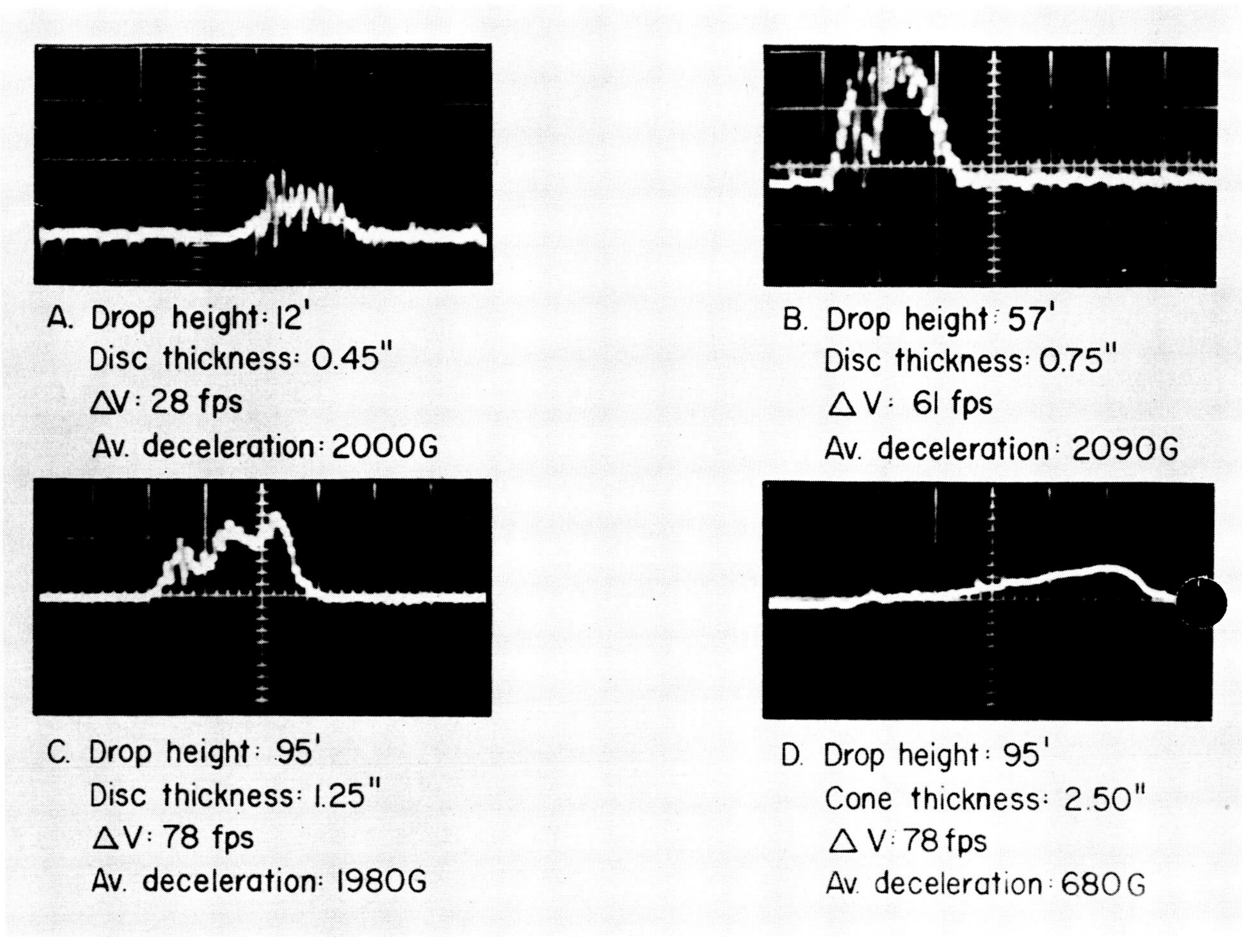

Figure 10

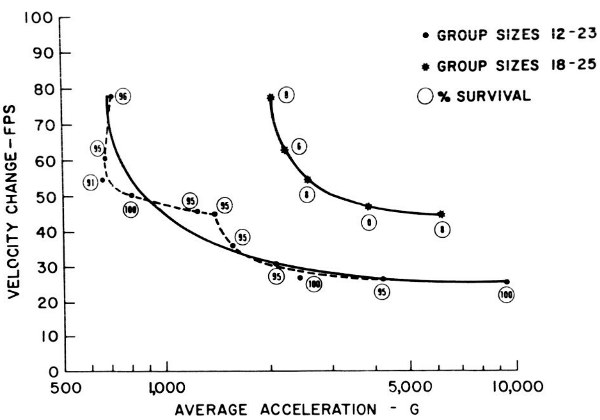

Figure 11

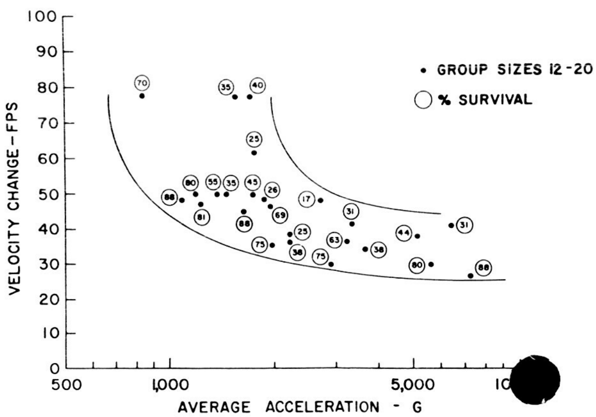

Figure 12

The tolerance curve obtained in this investigation (Figure 11) establishes for restrained mice the limits of transversely applied impact forces: a velocity change of 27 feet/second and an average deceleration of 650 G. The area below the curve constitutes a zone in which the probability of survival is extremely high. Zones of

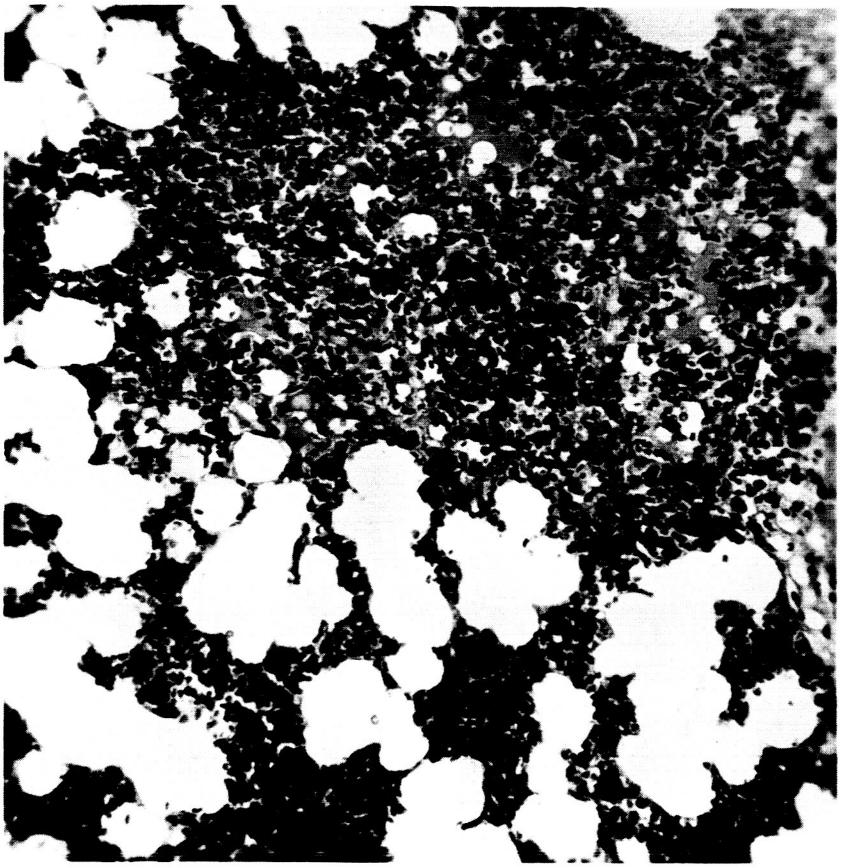

Figure 13

intermediate and extremely low probabilities of survival are also defined. Experimental evidence substantiates these conclusions to the extent of 78 feet/second and 9,300 G for the high-tolerance curve and 78 feet/second and 6,000 G for the low-tolerance curve. It is reasonable to assume, therefore, that this experimental approach to establishing criteria for protective-equipment design is feasible.

The solid curve drawn in Figure 11 was faired smoothly through the test points, although the dotted curve might be representative of the actual limit of tolerance. Two mechanisms for this discontinuous dotted curve suggest themselves. First, there might have been two modes of failure, as shown in Figure 5. Pathological examination, however, indicates one principal mode of failure. Second, Figure 4 demonstrates that application of different input pulse shapes can result in a discontinuous curve if all results are presented in one plot. For example, using the slow-rise and abrupt-fall triangular shape for the vertical-asymptote portion of the test series and then switching to a half-sine input shape for the horizontal asymptote would result theoretically (Figure 4) in a sensitivity curve very similar to the dotted curve in Figure 11. In point of fact, the four test points at the left of Figure 11 (below 1,000 G) were obtained with lead cones which produced a triangular input pulse shape (see Figure 10 D), and the right-hand portion of the curve was produced with lead disks which produced rough half-sine shapes (see Figures 10A, B, C). It is believed that this latter concept of different input pulses was responsible for the discontinuity, but this point will bear further experimental investigation.

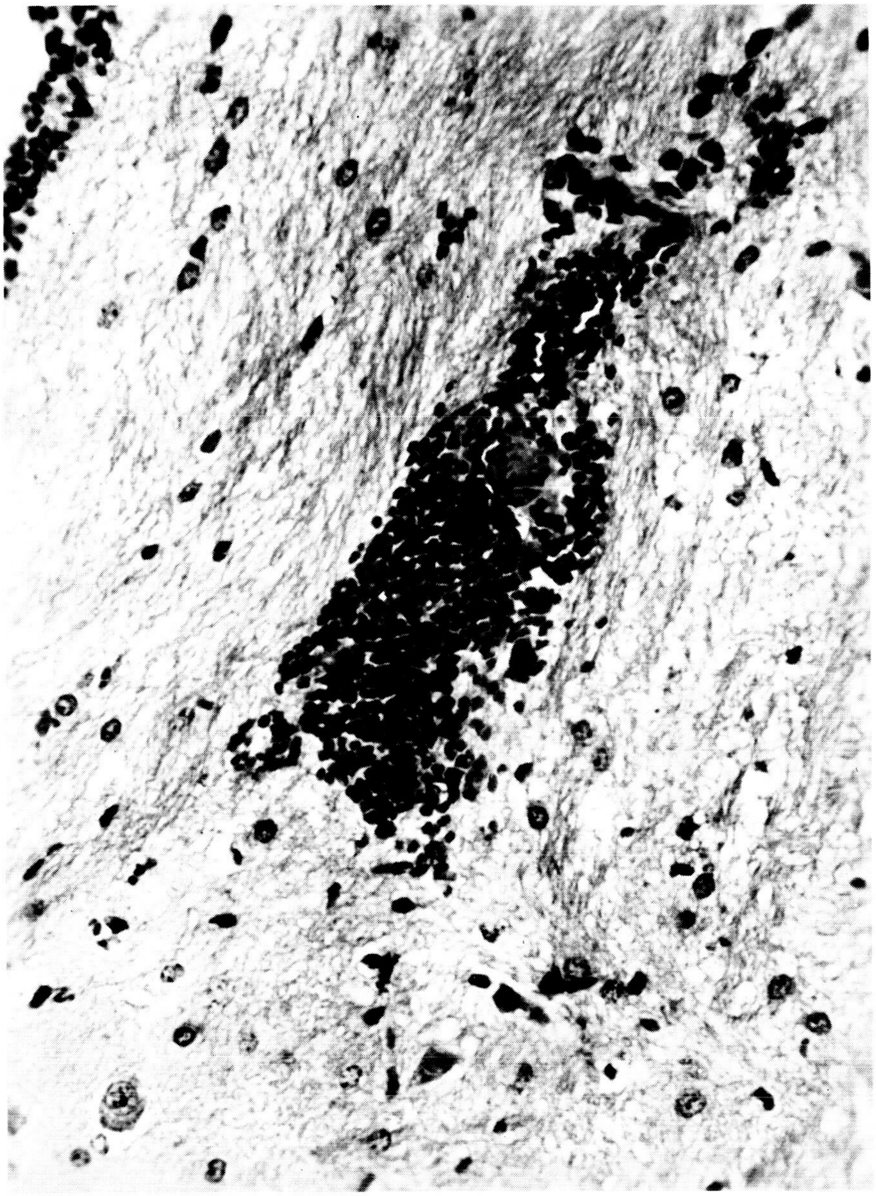

Figure 14

The Implications of the Mouse Results

Besides establishing by experimentation that one mammal, the mouse, has an impact tolerance which can be described adequately by the sensitivity-curve model, the mouse data also permit broad conclusions when compared with results for other animals. Conclusions may be drawn in regard to individual variations in g-tolerance, effects of differences in species (and size) on tolerance to impulsive velocity changes, and effects of differences in species (and size) on tolerance to long-duration accelerations.

First, in regard to variability in a population of test subjects, Figure 11 indicates a wide spread in g-tolerance of a genetically pure strain of male mice of similar

age, tested under controlled laboratory conditions. It may therefore be inferred that an even greater scatter for human response data should be expected.*

Figure 15 compares g-tolerance sensitivity curves for man and mouse. It is of interest to consider the differences between man and mouse vertical and horizontal asymptotes, in light of theoretical predictions of the mechanical model and experimental data provided by other animals.

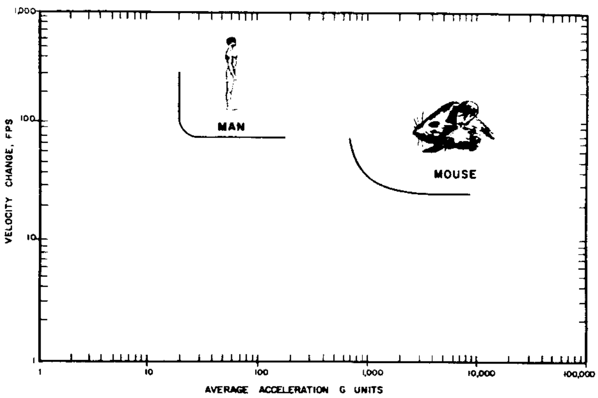

Figure 15. Impact sensitivity curves.

Theoretical considerations[1] indicate that geometrically similar structures of the same material but with different sizes will have identical horizontal asymptotes. Without consideration of how these restrictions are violated by comparison of different species, Figure 16 presents the data available for men[2], hogs[7], cats[8], and mice. The comparison of low-incidence-of-mortality curves shows a spread from about 27 to 80 fps. It appears that the impulsive velocity change required for irreversible damage to mammals is relatively insensitive to differences in species or size. Table 1, which includes free-fall data for unrestrained animals[9], further illustrates this point.

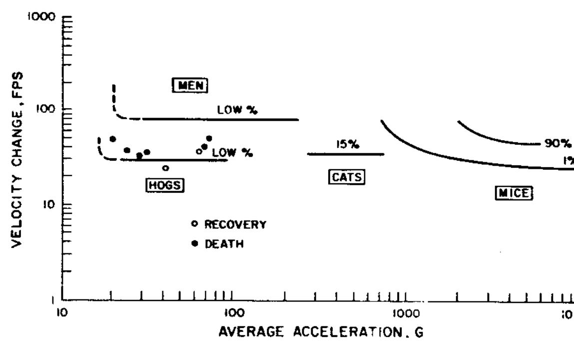

Figure 16. Mortality in the short-duration region.

In the matter of the long-duration g-tolerance asymptote, however, size is a very significant parameter. This is borne out by the theoretical prediction, that tolerance is inversely proportional to the first power of size and by the test data of Figures 17[10, 11] and 18.** The question concerning the desirable size of astronauts then arises. Figure 18 indicates a difference in g-tolerance of approximately 20 per cent between a five-foot man and a six-foot man which one would expect to be masked completely by individual variability in tolerance. This type of behavior was noted in the mouse tests, with statistical analysis of results for mice in the range of nine to 22 grams demonstrating little correlation of g-tolerance and size.

The conclusion is reached, as a result of the extensive mouse-impact test program, that the mechanical sensitivity model is indeed adequate for expressing g-tolerance data of mammals. Further, it is seen that this theoretical model leads to size implications that are reflected reasonably well by experimental data on mammals,

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*Note that Captain Beeding's famous sled test involved a velocity change of only about 48 fps, well below the 80 fps asymptote of Figure 6, and he suffered severe (but reversible) reactions.

**The Brachiosaurus is discussed in reference 10, in terms of estimates of weight and the strength of the fossil leg bones. Although not more precise than, perhaps, a factor of two, this estimate is sufficiently accurate for the rough comparisons of Figure 18.

modified considerably by differences between individuals in the "identical" population of test subjects.

TABLE 1

Mortality in the Short-Duration Region

Animal	Source	Velocity Change, FPS Threshold	LD50
Man	(2) Accidental Falls	80	--
	(9) Auto. Statistics	--	34
	(9) Estimated	21	26
Hogs	(7) Restrained	30	--
Cats	(8) Restrained	~30	--
Rabbits	(9) Unrestrained	~29	[illegible]
Guinea Pigs	(9) Unrestrained	~28	31
Mice	(6) Restrained	27	--
	(9) Unrestrained	~32	39
Rats	(9) Unrestrained	~37	44

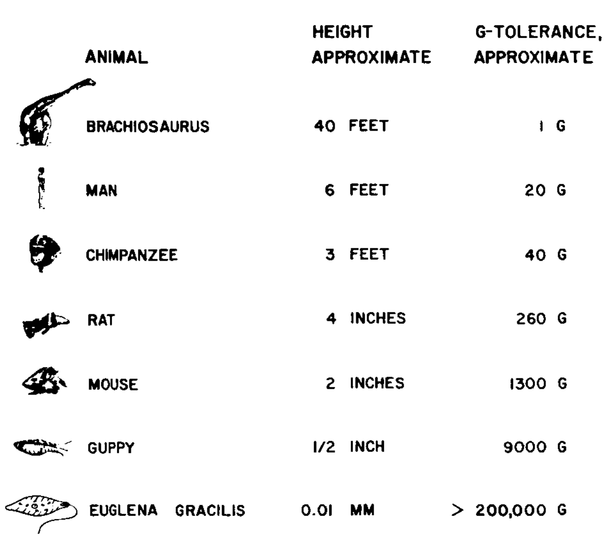

Figure 17

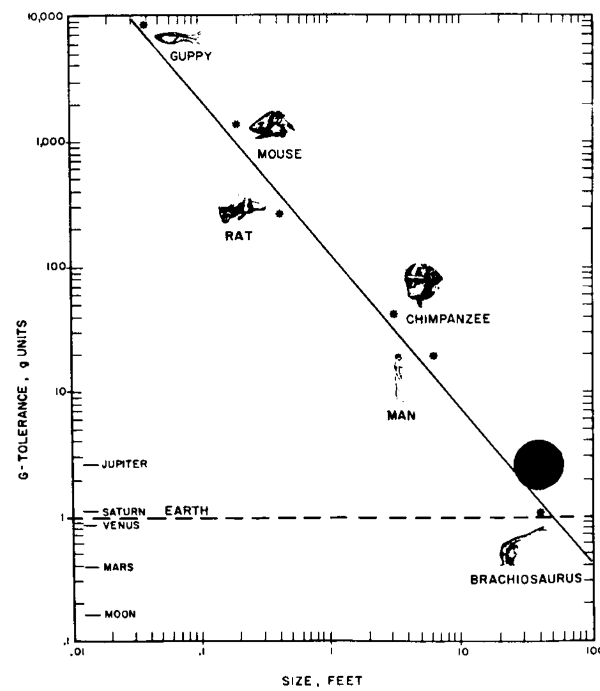

Figure 18. G-tolerance vs. size.

Symbols

g	Acceleration of gravity, 32.2 ft/sec^2
G	Applied acceleration, g units
Gav.	Time-average acceleration, g units
G_o	Average steady acceleration just sufficient to induce a given response
t_1	Duration of impact, or rise time of a pulse
T_n	Natural period of vibration
V_o	Velocity change required to produce a given amplitude of response
ΔV	Velocity change on impact

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AUTOMOTIVE HUMAN CRASH STUDIES

James J. Ryan
University of Minnesota

The kind of data most urgently needed to provide optimum safety for personnel subjected to impacts is basic engineering data. Application of fundamental principles to crash analysis is also required. Most engineering talent is being used to test out present designs, without the opportunity to devise and invent the energy-absorbing structures which greatly reduce impact forces. The research required to provide the necessary data for designers must be done by the design engineers themselves and, with them, by competent engineers familiar with the theory and practice of impact.

At the University of Minnesota, a 2,000-lb cart capable of speeds up to 40 miles per hour can carry dummy or human loads supported on forward or backward facing seats (or couches) to solve design problems in the absorption of energy and the reduction of impact forces. Graduate student personnel are available, although the problems as applied to human subjects require additional professional specialists. Complete instrumentation and control facilities allow implementation of impact studies. With equipment, the basic engineering data can be produced to determine the fundamental principles referred to above.

The Automotive Safety Research Project has carried on a program for several years at the Department of Mechanical Engineering of the University of Minnesota, under a research contract from the U. S. Air Force directed by Colonel John Paul Stapp (MC), Aerospace Medical Laboratory, then at Holloman Missile Center, N. M., (1955-1958), and a Grant-in-Aid from the U. S. Public Health Service, Bethesda, Md., (1959-1961). Under this program two reports have been prepared on "Safety Devices for Ground Vehicles," dated July 31, 1958, and September 1, 1960, respectively. Several papers were presented in 1960 and 1961 (see Bibliography).

It was the purpose of this Research Project to invent and test mechanical designs to reduce the destructive forces of collision on automotive and airspace passengers. The developments which have been produced in this laboratory include engineered automatic seat belts, hydraulic shock-absorbing bumpers, a large padded steering post with a short-travel absorber and a retracting steering-wheel rim for the driver, and a dash recessed under the windshield in front of the passenger. With judicial padding, the flailing of the arms and the legs of the body held by the seat belts are further protected from injuries.

Tests have been made in a car with a driver and passenger safely impacting a rigid barricade at 20 mph (Figure 1); stopping the cart with a bumper at 18.75 mph having a free-swinging live passenger on a seat belt (Figure 2); and having a seat-belted human safely decelerate on the cart with a bumper from 25 mph, aided by a damped, collapsible steering wheel with a large central pad (Figure 3). From the instrumentation and photographic records of these and other tests, design data have been acquired which would make possible a reduction of human-impact forces to

Figure 1. Passenger car impact at 20 miles per hour showing driver and passenger with seat-belts at Air Force Missile Development Center, N. M., November 13, 1957.

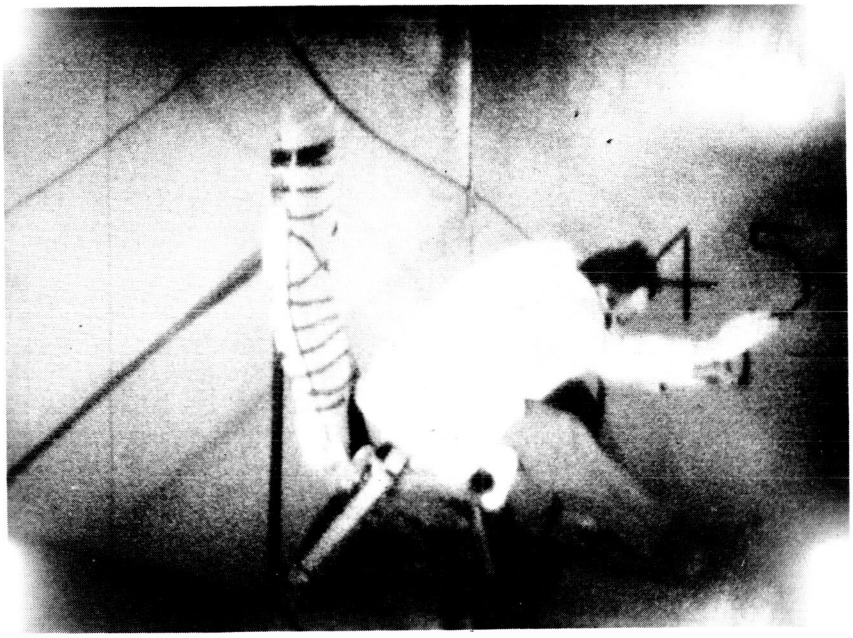

Figure 2. Impact displacement for human crash tests, with Peter Schoeck riding cart at 18.5 mph velocity.

one-quarter, with an estimated reduction of injuries and deaths in automotive accidents to one-half.

Figure 3. Impact displacement for human crash test, with Peter Schoeck riding cart at 24.85 mph velocity with padded steering wheel.

Deceleration Test Cart

A research facility was constructed in the form of a 2,000-lb steel cart (one-half the weight of an automobile) which, mounted on rails and propelled by a ten-inch air gun, obtained speeds from 15 to 40 mph. This cart was decelerated by an automotive hydraulic shock-absorbing bumper capable of applying a relatively uniform force over the stopping distance to absorb the maximum energy with the minimum force. An outline drawing of the cart is shown as Figure 4.

The first tests with the cart were made to determine the characteristics of the hydraulic shock-absorbing bumper (Figure 5) by changes in the design of the metering pin. Oscillographic recordings of hydraulic bumper internal pressure, cylinder stress, and piston displacement were made for a series of velocity impacts from 15 to 40 mph, and for approximate cart loads of 1,500, 2,000, and 2,500 lbs. At the same time, dummy riders were placed on the cart and oscillographic and photographic measurements of deceleration, seat-belt pull, dummy displacements and dummy decelerations were made. After a suitable hydraulic absorber was obtained, seat-belt tests with dummies and humans were continued to determine the minimum seat-belt forces with engineering control. It was found that hydraulic seat-belt tighteners not only removed the slack in the belts but held the body firmly in the seat to reduce the forces to a desirable minimum. Positioning the body also made possible the use of a large padded steering post situated properly to fit the chest as the upper torso rotated forward on the seat belt. The rider beside the driver (in an automotive vehicle with a recessed dash) also was benefited by the seat-belt tightener in that better body control allowed more positive protection for the upper torso, the head, and the flailing extremities.

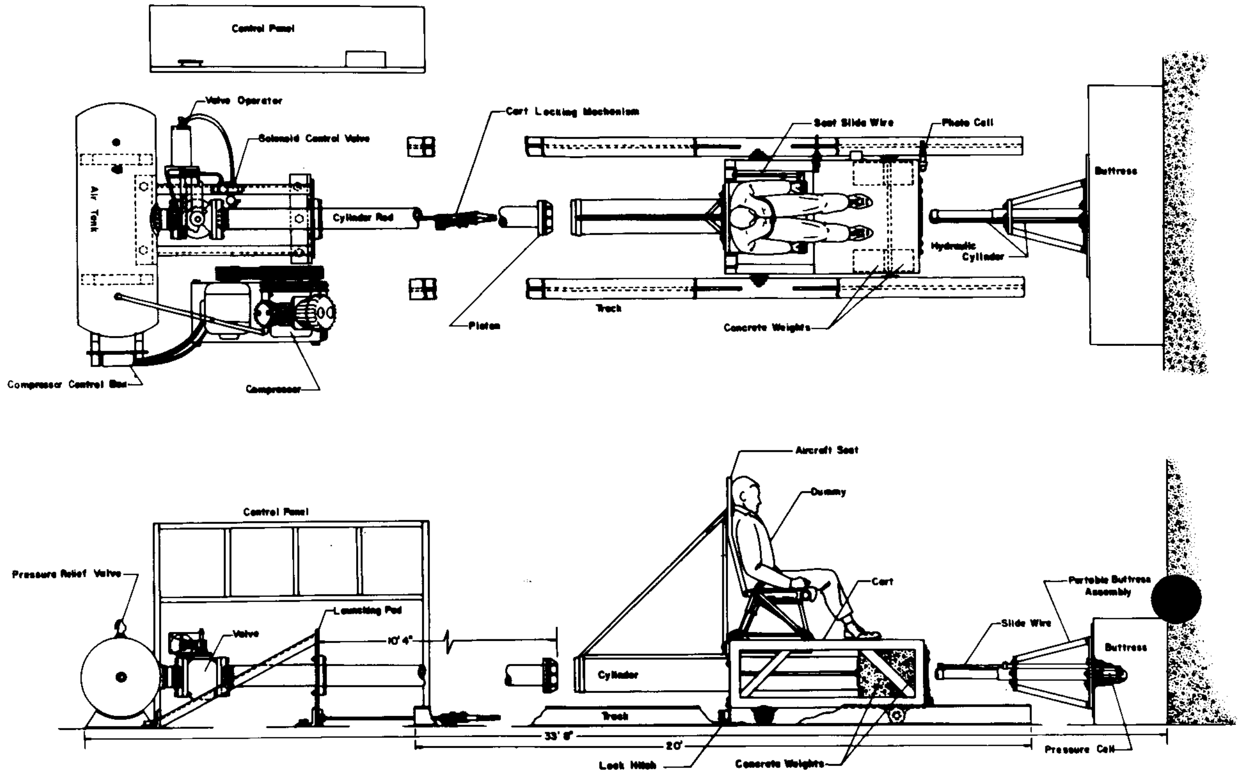

Figure 4. Schematic diagram of 2,000 lb cart, driven by air gun, with dummy on seat-belts and hydraulic shock-absorbing bumpers on wall.

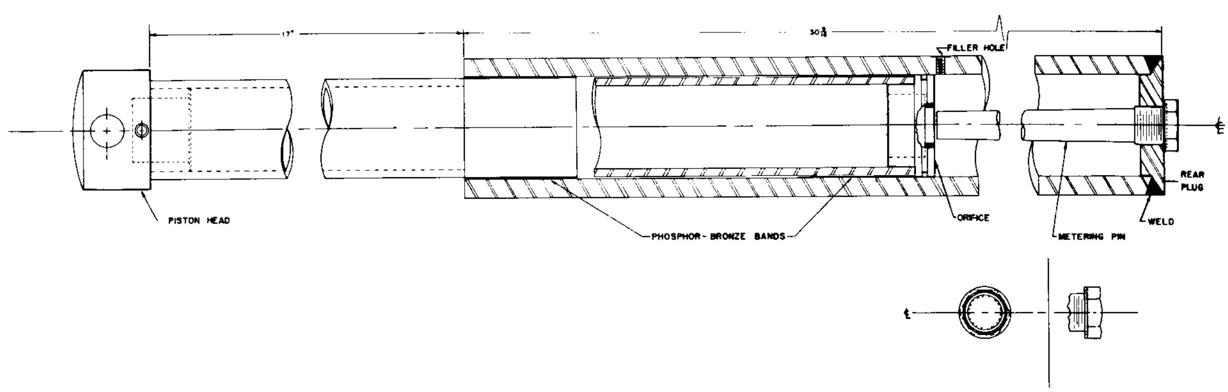

Figure 5. Outline drawing of hydraulic shock-absorbing bumpers for passenger car—2-3/4 in. Piston diameter—1-in. orifice.

The final design of the bumper, which was generally acceptable for automotive application, gave measurements of deceleration and time for a constant cart weight of 2,100 lbs for various speeds in the 20-mph range as shown on Figure 6. These are impulse curves, given as A, B, and C, and their area is proportional to the product of mass and initial velocity. To determine the energy absorption of the bumper on the cart, these curves are plotted for deceleration versus displacement as shown in Figure 6 as curves D, E, and F. It is observed that the forces are not uniform for the displacement, although the uniformity of forces over the major portion of the displacement is acceptable as shown by the horizontal lines, G, H, and I, superimposed upon

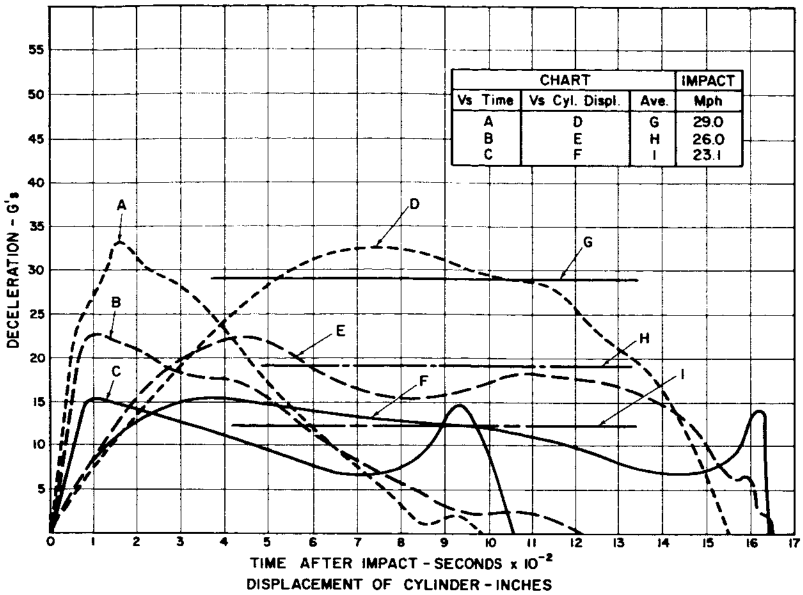

Figure 6. Cart Deceleration for constant cart weight of 2,100 lbs at various impact velocities with hydraulic bumper.

the force curves. The theory that the force resistance of an orifice controlled by a metering pin may approach proportionality with the velocity was generally supported.

Using the same metering-pin design, tests were made to determine the deceleration response of the cart to various weights at a constant speed of 29 mph. Figure 7 shows the deceleration curves with respect to time and distance for three runs with

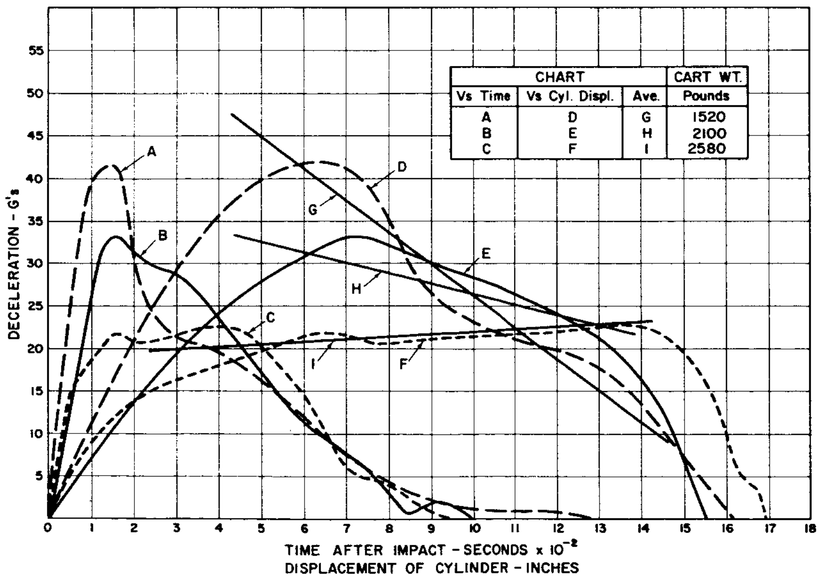

Figure 7. Cart deceleration for constant impact velocity at 29 mph with various cart weights with hydraulic bumper.

1,520, 2,100, and 2,580 lbs as the weights of the cart. With curves D, E, and F presenting the outline of the force-distance energy curves, the slope of the force decay would be indicated by the lines G, H, and I. Thus, for a given metering pin, the lightest weight of the car would give the earliest maximum forces, while with an extremely heavy load, the vehicle would tend to have increasing forces at the end of the stroke.

It was originally thought that the automotive hydraulic shock-absorbing bumper should have a constant force of deceleration to reduce to the minimum the forces imposed upon the person on an elastic seat belt. As shown in Figures 8 and 9, earlier application of force is desirable as a means of reducing the maximum forces on the person with a seat belt. Further, for any given damping in series, the seat-belt forces may be determined as shown in these figures.

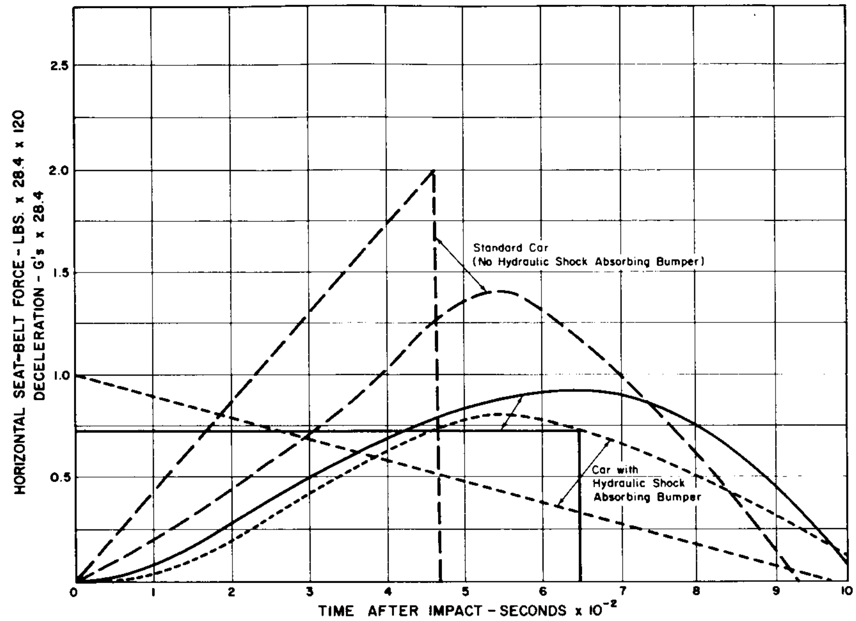

Figure 8. Hypothetical cart decelerations with rectangular and triangular bumper characteristics at 30 mph and corresponding horizontal seat-belt forces on man with equivalent weight of 120 lbs. with 62.5% critical damping. (200 lb. man) Decel. Distance, 17 in. Seat Belt Nat'l

Theoretical and experimental design data for hydraulic shock-abosrbing bumpers have been correlated for a 2,000-lb vehicle. The theoretical equation for the area of the orifice, A_o in^2, is

$$A_o = \frac{V}{c} A_p^{\frac{3}{2}} \sqrt{\frac{d}{2Wa}}$$

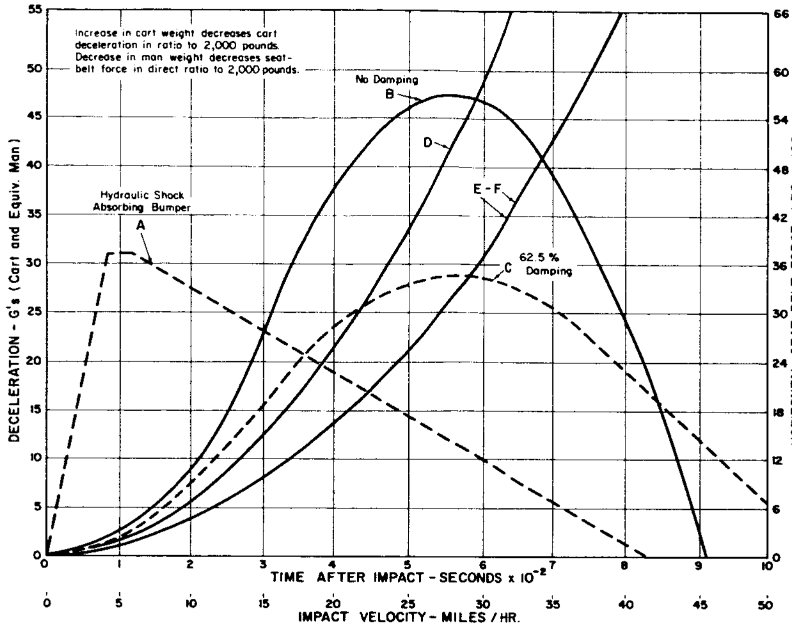

Figure 9. Hypothetical cart deceleration (A) with horizontal seat-belt force without damping (B) and with 62.5% critical damping (C) at 30 mph; and summary of maximum seat-belt forces from 0 to 40 mph on man with equivalent weight of 120 lbs. without (D) and with (E) damping. Maximum cart deceleration (F) from 0 to 40 mph.

where $\underline{V}$ is the velocity of the piston, in/sec

$\underline{A_p}$ is the area of the piston, in^2

$\underline{d}$ is the density of the fluid, lbs/in^3

$\underline{W}$ is the total weight of the cart or vehicle

$\underline{a}$ is the deceleration of the piston, in/sec^2

and $\underline{c}$ is the orifice discharge coefficient.

For the automotive-type deceleration, the diameter of the orifice enclosing the metering pin is one inch, and the maximum stroke of the piston is 17 inches. The values of the orifice discharge coefficients vary from 0.60 to 0.70, with an average of 0.[illegible] The impact conditions for the first inch of travel are not shown, and the orifice is sealed off for the last one-half inch. From the experimental measurements, relationships have been obtained between the pressure, the velocity, and the orifice area, and a correlation has been established with the theoretical data for application to other deceleration conditions.

Figure 10 shows the necessity for a seat-belt tightener as part of the hydraulic-absorber assembly to properly place the individual for maximum seat-belt effectiveness. Figure 11 shows the forces and acceleration which would be tolerable to a human with a seat belt at 40 mph impact with the proper mechanical system.

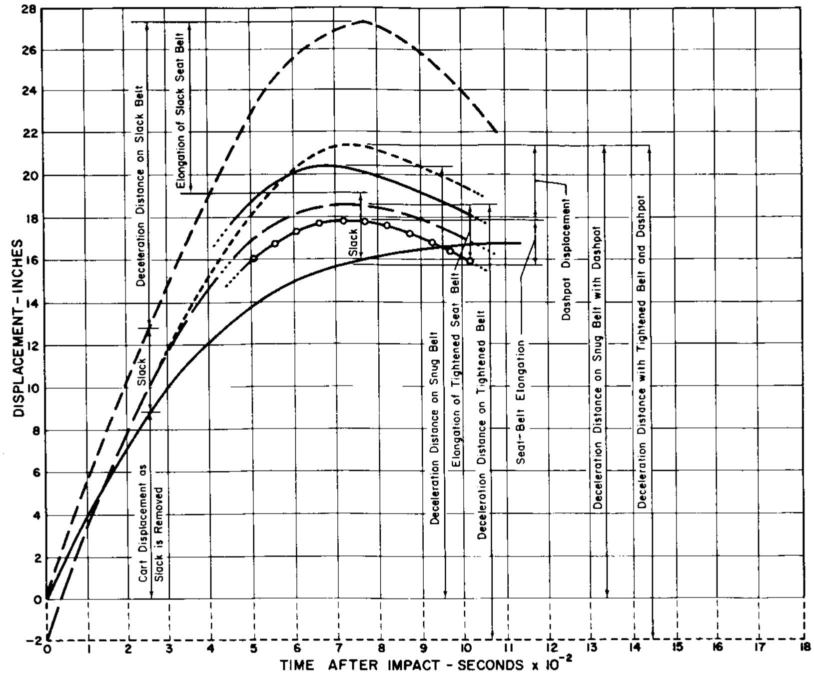

Figure 10. Relative deceleration displacements of hydraulic shock absorbing bumper on cart at 25 mph for an equivalent weight of 120 lbs., (200 lb. man) on slack, snug, tightened and series-damped seat belts (Slack includes Seat Movement).

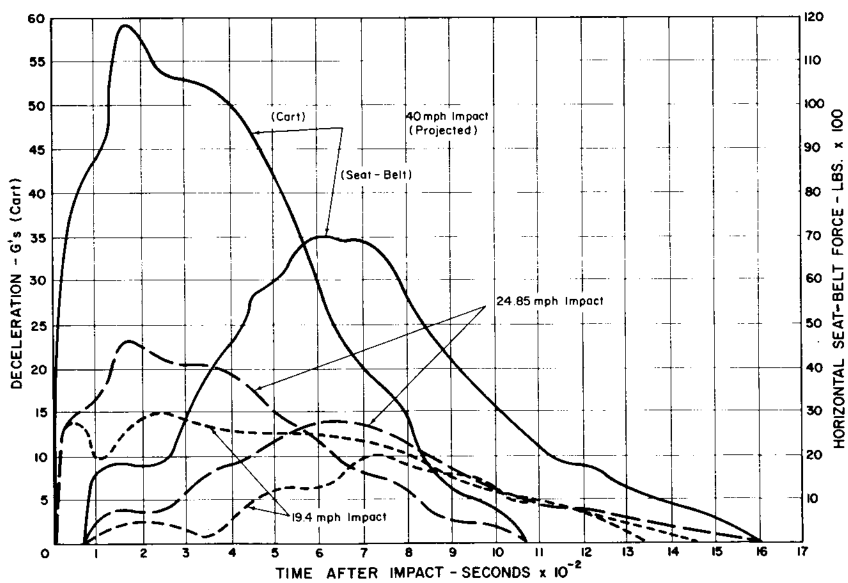

Figure 11. Test cart deceleration and seat-belt forces on Peter A. Schoeck (185 lbs) at 19.45 and 24.85 mph with steering wheel having 3 in. travel and single belt with tightener—Projection to 40 mph impact. (Total seat-belt forces 15% greater.)

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Automotive Safety Research Project

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2. Ryan, J. J. and BeVier, W. E.: "Safety Devices for Ground Vehicles," U.S.P.H.S. Report, September 1, 1960, 217 pages.

3. Ryan, James J.: "Human Crash Deceleration Tests on Seat Belts," presented before the Annual Meeting of the Aerospace Medical Association, Chicago, Illinois, April 26, 1961. Aerospace Medicine, February 1962, Vol. 33, No. 2, pp. 167-174.

4. Ryan, James J.: "Reduction of Crash Forces," presented before the Fifth Stapp Car Crash and Field Demonstration Conference, September 14, 1961, at the University of Minnesota. Published in a book of the Proceedings of the Fifth Stapp Conference, by the University of Minnesota Printing Department, March 1962, as Chapter V, pp. 48-89.

5. Ryan, James J.: "Mechanical Reduction of Impact Forces by Automotive Design," presented before the annual meeting of the American Medical Association, New York City, New York, June 27, 1961, at the Symposium on Medical Aspects of Automobile Injuries and Deaths. Published in Research Review, a quarterly journal of TRAFFIC SAFETY Research, a publication of the National Safety Council, Chicago, Illinois, Vol. 6, No. 2, June 1962, pp. 3-14.

6. Ryan, James J.: "Crash Deceleration Tests with Human Subjects," presented before the Fourth Annual Meeting of the Human Factors Society, Boston, Massachusetts, on September 14, 1960. Published by McGraw-Hill Book Company in the book "Human Factors in Technology" as part of the proceedings of the symposium including "New Approaches to Highway Safety" in August, 1962, Chapter 15.

12875

INVESTIGATIONS ON LONG-STRETCHING BODY RESTRAINTS

Dr. Bertil Aldman
Bromma, Sweden

It has been demonstrated in several experimental car collisions that there is a time delay between the deceleration of the restrained dummy relative to the car. The time delay was verified for European cars in an experimental series of car-barrier impacts in Sweden in 1958.

Figure 1 shows that most of the deceleration of the car is over before the deceleration of the dummy stops. (The velocity is given in meters per second, the time in milliseconds.) This time delay can only partly be explained by a supposed slack in the

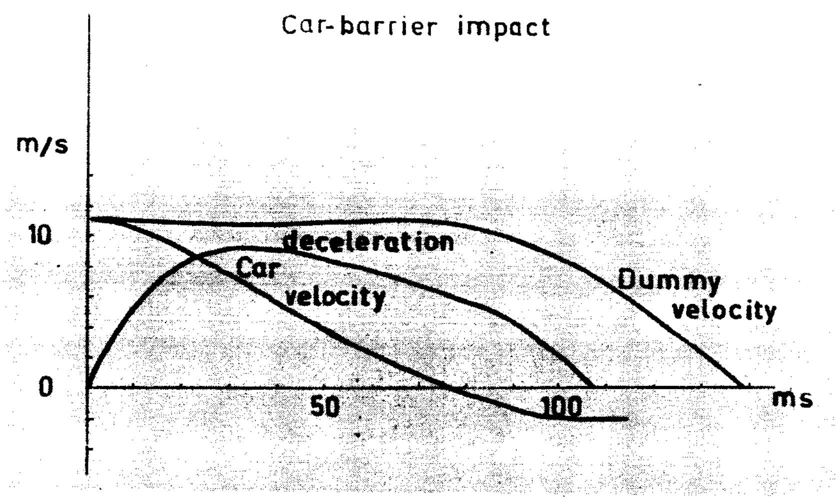

Figure 1. Car velocity and deceleration and dummy velocity following barrier impact. Note that the dummy velocity continues after the car stops.

restraint. Also the elasticity of the car and the restrained body is of importance. If we want the restrained body to take part in the deceleration of the car, the time delay is most undesirable. As there is a great difference in velocity between the anchoring points for the straps and the restrained body, the rate-of-onset of deceleration might be very high if we chose a short-stretching strap as a restraint. A possible way out of that problem is to change the deceleration pattern of the vehicle.

In my investigations I have chosen another way which is necessary if the space in the vehicle is so limited that upper-torso restraint is desired, and that is to use long-stretching straps for the restraint. The use of such straps means that even with very high rate-of-onset of deceleration for the vehicle, the rate-of-onset of deceleration

for the restrained body will be governed by the stretching properties of the straps. Neither the time delay nor the deceleration pattern of the vehicle will be of much influence if the stretching properties of the restraint are such that the deceleration of the occupant is tolerable under the worst possible conditions. Such a restraint, therefore, can be used even in vehicles of very rigid construction. Automobile safety belts of such construction have now been used in Sweden for more than three and a half years, and today about 60 per cent of Swedish cars are equipped with approved belts, tested to give maximal deceleration for the occupant of about 50 G.

It has been possible to prove a good effect of these belts in automobile accidents. Bäckström[1] and Lindgren have reported between 70 and 90 per cent protection against dangerous-to-fatal injuries for persons using belts in accidents, and we have found no cases with dangerous injuries caused by the belts, where these have been properly used. The principle for reducing the rate-of-onset by the stretching of the straps is illustrated in the Figure 2. It shows the velocity-time patterns and the deceleration-time patterns for the anchoring point (1 on the graph) and three different points of the single-strap length during impact. It also shows the corresponding deceleration time patterns, in meters per second squared. Again, curve 1 corresponds to the attaching point of the vehicle, and curves 2, 3, and 4 are for other points on the belt. You can see here how the rate-of-onset of deceleration is changed for the different points, only because of the stretching of the straps. The maximal elongation of the straps during the impact was 24 per cent of the original length. For a complete belt quite a lot of factors would influence the rate-of-onset; for instance, strap length, direction, and number of straps, type of seat, etc. which you can see in the film to be shown later. That makes the rate-of-onset for the complete assembly much lower than that of the seat. Even for straight lengths of a single strap, the elongation is not uniform during dynamic loading, but a complicated system of strain-waves (Figure 3). That phenomenon shows that there are two possible ways for breaking the strap. If we overload the strap—by using a too-heavy weight, for instance—the force may break the strap near zero velocity and the strap has taken up all the energy it possibly can. Alternately if we try to increase the stretching velocity too much, for instance by using too high a rate-of-onset of deceleration, it may cause breakage due to a local strain wave at a time when the strap hasn't taken up much energy. In order to avoid that in a complete assembly, the safety belts should be tested dynamically with a considerably higher rate-of-onset, so we can

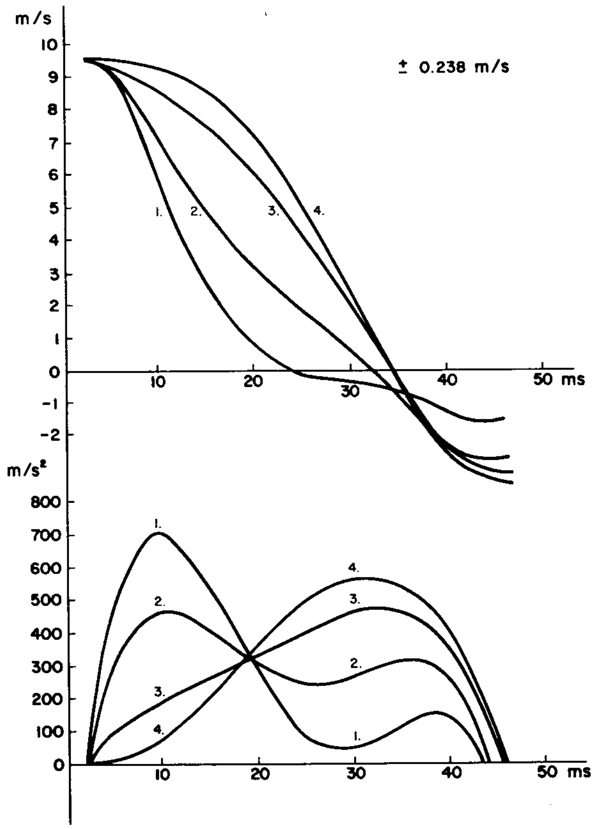

Figure 2. Dynamic stretching of single strap. Velocities and deceleration of point on a restraint strap during impact. 1. Near the anchoring point in the test vehicle. 2. In the middle of the strap. 3. Near the loading weight. 4. At the loading weight.

be sure that if breakage occurs it should be by overloading, so that the assembly will take up all the energy it possibly can.

In reducing the rate-of-onset of the deceleration of the restrained body, we increase the possibility of protecting the skeleton from damage; but what will happen to internal organs? They will only indirectly be decelerated by the safety belt, as they will have a possibility of moving a certain distance before they are decelerated by their anchors to the body or by its limitation. In other words, they will have their own deceleration pattern. It is therefore very difficult to find one single figure for the deceleration or the rate-of-onset of deceleration of the whole body. In order to study the displacement of different internal organs during acceleration, I made an experimental series using pigs as test animals. The pig was placed on an oscillating table. Contrast dye was injected in the aorta. In Figure 4 you can see the heart pressed up to the top of the chest cavity, then at the other end of the stroke, there is quite a lot of space on top of the heart. These marked displacements of internal organs during the deceleration along the long axis of the body speak in favor of decelerating a human body in an upright seated position, so that the deceleration will take place at about right angles to the long axis. A deceleration in that position makes upper-torso restraint necessary, and in order to avoid chest or neck injuries from such a restraint the rate-of-onset must be low enough for the force to be as evenly distributed over the chest as possible. That can be done very easily by using long-stretching straps for the upper-torso restraint. We have found it necessary to have long-stretching straps for an upper-torso restraint, because we have seen some accidents in which short-stretching straps were used and the ribs broke all along the edge of the strap. Figure 5 shows a car that hit a huge tree at a speed of about 80 miles per hour. The driver was killed by the steering column, which comes up almost to the roof of the car. The right-hand front-seat passenger sustained very bad leg injuries because the engine was compressed up to the front of the seat, but he had no chest injuries and no head injuries. He stayed conscious during the whole impact and could tell what had happened. He was using an upper-torso restraint of the long-stretching type. We have seen several very violent accidents where passengers obviously have been saved by using safety belts of that type.

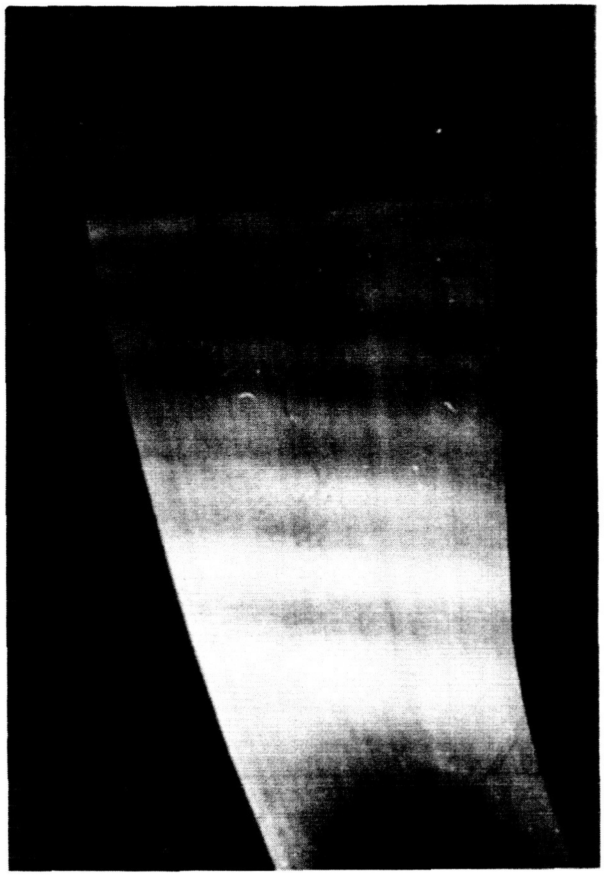

Figure 3. Strain waves passing the strap several times during elongating in dynamic testing. The drum camera film has moved in a vertical plane parallel to the moving direction of the strap (which in the picture is from right to left). When the hindmost anchorage (to the right) stops, a strain wave (the dark oblique line) starts, reaching the foremost anchorage after one m/sec. The next strain wave starts two m/sec after the first one, etc.

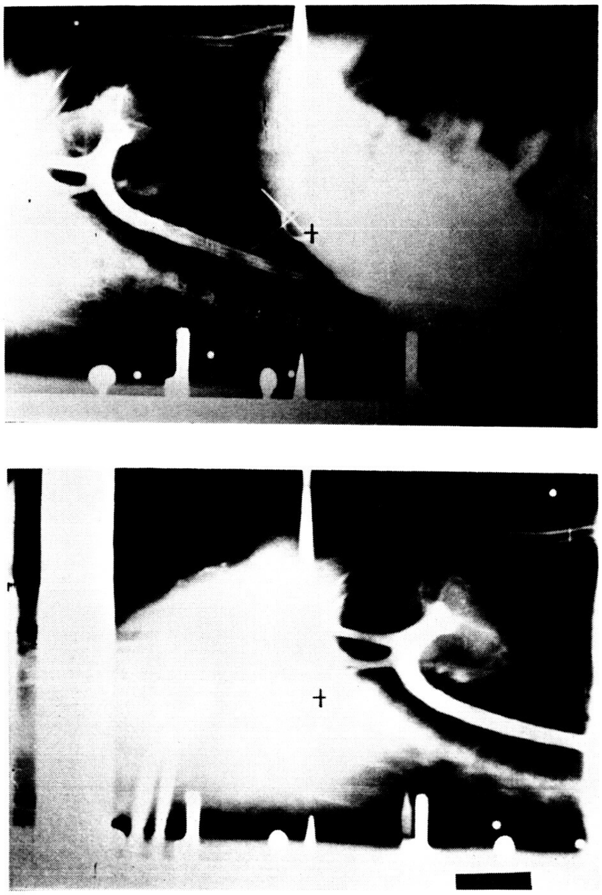

Figure 4. A pig during horizontal vibration, with contrast dye in the blood stream. Top: The heart impacts the anterior of the thorax, folding the left subclavian artery at almost 90 degrees. Bottom: The heart inertially swings posteriorly in the thorax, stretching the left subclavian artery, a not unusual site of hemorrhage following impact. Note also the altered angles to the aorta of the intercostal arteries.

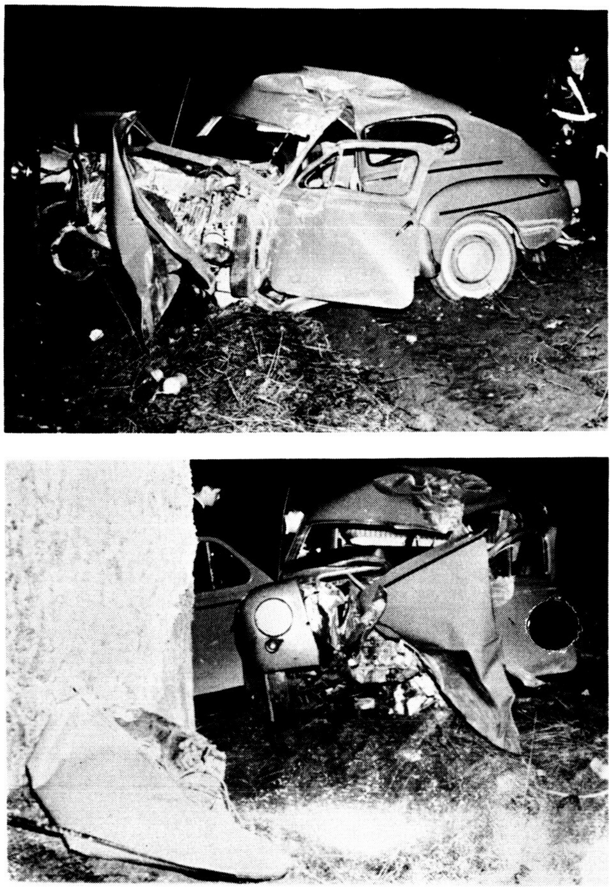

Figure 5. Results of a tree collision at 80 miles per hour. The driver was killed on the steering column. The right-hand passenger, wearing a lap belt and long stretching diagonal chest strap, had severe leg injuries due to the engine penetrating to the front seat, but had no head or chest injuries, and remained conscious after the impact.

(A film was shown dealing with dynamic testing of various types of car safety belts.)

In those impact experiments we had used a bucket seat with a tiltable backrest. That is in order to save the cost of repairing the seat-back, because it always breaks on impact. The force necessary to break the backrest of an ordinary seat is about 200 to 300 pounds, readily exceeded by its own inertia, or especially if the upper-torso restraint is attached to the seat backs during even a moderate impact. A half-car body was used in these experiments, so the dummy could slide forward more than it would in an actual car. These tests established that for the small car the best restraint of five tested has a lap belt and a long stretching diagonal chest strap attached to the door post and joining the lap belt in back of the hip[2].

Question: What onset rate was used?

Aldman: The rate of onset of deceleration is very high. The braking distance is about 2-1/2 inch in this vehicle, and that is the standard test for approval of belts in Sweden. We use exactly that equipment for standard testing, and for the reasons I told you before we like to test them with an extremely high rate of onset of the deceleration. Of course it's not actually the same situation in a car at that speed but at a higher speed you may have a very high rate of deceleration. For instance in an 80-mile-per-hour collision with a tree that you saw in Figure 5 I think there must be a very high rate of onset of deceleration of the car in that case. However, we have seen that it is very dangerous to hit a fixed obstacle at the speed of about 25 miles per hour, and at even less speed you may have fatal injuries.

Hasbrook: What was the device or mechanism that you used to accelerate the car and what was the acceleration rate?

Aldman: The car was accelerated by a falling weight at about 1-1/2 G.

Question: I was wondering about the characteristics of material in your belt. Does it flow or elastically stretch?

Aldman: The material is a dacron webbing in the straps. Both a yielding and elastic rebound occurs on impact. Of course in the human body we have some elasticity which will cause the rebound effect of the human even if there is no rebound effect in the straps.

Question (to Dr. Perey): Can you detect end plate fractures by x-ray without contrast dye injected into the nucleus? We have done some of this work and generally could not detect the end plate fracture. I think out of 28 cases we had four that we could pick up on the x-rays. Otherwise we had to dissect in order to determine the fracture. We knew the end plate had failed but we couldn't pick it up on the x-ray.

Perey: It is very difficult to see these fractures in living people.

Question: I have another question along that line or perhaps it is a comment. This is in regard to the work Professor Patrick was talking about too. Actually the loads we got to produce these initial end plate fractures were about half as much I believe as the loads you got but the fractures we got couldn't be demonstrated unless we actually took the disc off the bone. We found the fracture by the blood coming through the fracture line—the fracture was so fine, such a hair line fracture. From the appearance of yours, apparently the additional load has produced the additional extension of the fracture. We wondered if you had found any of these very very fine hairline fractures. Now unfortunately due to experiences beyond our control, we couldn't use cadaver material that was younger than 60 years old. We noticed that in your tests the loads on individuals older than 60 were considerably less than on younger individuals, but still to produce these very fine initial hairline fractures, our loads were about half as large as yours.

Perey: Yes, I have seen that and in the reprint I put out there on page 55 you can see the microscopic picture with a very fine fracture[3].

Question: I have a question for Dr. Aldman concerning the belt material. I'd read some time ago that Sweden had been doing research in steel fibers in their belts and I may have missed this—was that a part of this research program? If so, what were the results?

Aldman: No it wasn't part of this program. The results were very discouraging and so they have stopped that.

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1. Bäckström, C. G.: Säkerhetsbalten i bilar. Nord. Med. 62:1667, 1959.

2. Aldman, Bertil: Biodynamic Studies on Impact Protection. Acta Physiologica Scandinavica, Vol. 56, Supplementum 192, 1962.

3. Perey, Olof: Fracture of the vertebral end-plate in the lumbar spine. An experimental biomechanical investigation. Acta Orthopaedica Scandinavica Supple. XXV, 1957.

APPENDIXES

APPENDIX I

Symposium on
IMPACT ACCELERATION STRESS

Under the Auspices of the
Man in Space Committee
of the
Space Science Board

November 27-29, 1961
Brooks Air Force Base
San Antonio, Texas

National Academy of Sciences–National Research Council
2101 Constitution Avenue, N.W.
Washington 25, D. C.

TUESDAY, November 28, 1961

Classroom A, School of Aerospace Medicine

MORNING SESSION

Review and Forecast of Impact Studies I

8:30 *Introduction*—JAMES D. HARDY

8:40 *Review of Swedish impact work and plans*—H. BJURSTEDT, O. PEREY, B. ALDMAN

9:10 *Review of Netherlands impact work and plans*—M. P. LANSBERG

9:40 *Review of British impact work and plans*—J. GUIGNARD

10:10 *Review of Canadian impact work and plans*—WILBUR R. FRANKS, H. B. HAY

10:40 *Review of French impact work and plans*—FRANCOIS VIOLETTE

11:10 *Review of Japanese impact work and plans*—M. OSHIMA, I. SAITO

AFTERNOON SESSION

Review and Forecast of Impact Studies II

1:30 *Aviation Crash Injury Research—Review of the Flight Safety Foundation impact work and plans*—MERWYN A. KRAFT

1:45 *Automobile Crash Injury Research—Review of the Cornell University impact work and plans*—ROBERT WOLF

2:00 *Impact studies of other United States universities*—H. R. LISSNER

2:30 *Impact studies of the United States automobile industry*—A. L. HAYNES

3:00 *Impact studies of the United States aerospace industry*—G. A. HOLCOMB

3:45 *Biomechanics of impact injury*—HENNING E. VON GIERKE

4:15 *The significance of "jerk"(G/sec) on injury production and mechanisms of physiological shock following impact*—J. P. STAPP

EVENING SESSION

7:30

Panel Discussion

Crash Impact Studies: Present Work and Problems

Chairman, H. R. LISSNER

C. GADD	W. LANGE	J. J. RYAN
A. L. HAYNES	A. MOSELEY	J. W. TURNBOW
D. F. HUELKE	G. J. PESMAN	R. WOLF

MONDAY, November 27, 1961

Classroom A, School of Aerospace Medicine

MORNING SESSION

9:00

Opening Address and Presentations of U.S. Government Agencies

Welcome DR. JAMES D. HARDY, *Chairman,* Yale University

Address BRIG. GENERAL THEODORE C. BEDWELL, JR., Commander, Aerospace Medical Division (AFSC), Brooks Air Force Base

9:30 *Review of NASA impact work and plans*—ALFRED L. MAYO, Deputy Director of Aerospace Medicine, Office of Manned Space Flight, NASA

10:00 *Review of USAF impact work and plans*—KARL H. HOUGHTON

10:30 *Review of USN impact work and plans*—F. K. SMITH

11:00 *Review of USA impact work and plans*—J. C. BEYER

11:30 *Review of FAA impact work and plans*—JAMES GODDARD

AFTERNOON SESSION

2:00

Tour of School of Aviation Medicine, Brooks Air Force Base

This symposium is supported by a grant from the National Aeronautics and Space Administration. Assistance was provided also by the Department of the Air Force and the Department of the Navy and their cooperation is gratefully acknowledged.

WEDNESDAY, November 29, 1961

Classroom A, School of Aerospace Medicine

MORNING SESSION

Review of Research Areas on Impact

8:30 *Experiences in head and skeletal injury research* — E. S. GURDJIAN

9:00 *Mechanics of acceleration concussion*—R. FRIEDE

9:30 *Damage to internal organs*—F. GAYNOR EVANS

10:30 *To what extent can animals instead of man be used in impact testing?*—J. D. MOSELY

10:50 *Comparison of the dynamic characteristics of dummies, animals and men*—ROLF R. COERMANN

11:10 *Application of the impact sensitivity method to animate structures*—M. KORNHAUSER, A. GOLD

11:30 *Characteristics of materials for absorbing energy of impact*—JOHN MOORE

AFTERNOON SESSION

1:30

Panel Discussion

Human Impact Studies: Present Work and Problems

Chairman, J. J. SWEARINGEN

E. BEEDING	A. E. HIRSCH	P. PAYNE
R. BESCO	G. A. HOLCOMB	J. P. STAPP

EVENING SESSION

7:30

PANEL DISCUSSION

Protection Against Impact, Including Water Immersion

Chairman, H. BLACKSHEAR

R. BESCO	H. E. FREEMAN	G. A. HOLCOMB
G. F. BOND	R. F. GRAY	P. PAYNE
R. L. CARTER		

END OF SYMPOSIUM

THURSDAY, November 30, 1961

Tour of Holloman Air Force Base, New Mexico

APPENDIX II

LIST OF PARTICIPANTS

Aldman, Dr. Bertil
Utflyktsvagen 36
Bromma, Sweden

Armstrong, Dr. Robert
Life Sciences Section
Convair Astronautics
San Diego, Calif.

Austin, Lt. Col. Richard B.
Chief, Aviation Branch
Plans Division, Plans, Supply
& Operations Directorate
Room 2815, Main Navy Building
Washington 25, D. C.

Barr, Dr. Norman L.
Head, Biological Sciences and
Systems Department
Defense System Division
General Motors Technical Center
Warren, Michigan

Bates, George P.
Aircraft Development Services
Federal Aviation Agency
Washington 25, D. C.

Bedwell, Gen. T. C., USAF MC
Commandant of the School of
Aerospace Medicine
Brooks Air Force Base
San Antonio, Texas

Beers, Dr. George R.
Holloman Air Force Base
New Mexico

Be[illegible]o, Robert
Aerospace Group
Research and Development Division
Hughes Aircraft Corp.
Culver City, Calif.

Beyer, Maj. James C., USA MC
Medical Research and Development
Command
Headquarters, Department of the Army
Office of the Surgeon General
Washington 25, D. C.

Bezreh, Capt. A. A.
U. S. Army Board for Aircraft
Accident Research
Fort Rucker, Ala.

Bond, CDR George F., USN MC
U. S. Naval Medical Research
Laboratory
U. S. Naval Submarine Base
New London, Groton, Conn.

Bommarito, C. L.
The Boeing Company
5144 S. Myrtle Street
Seattle 18, Washington

Brian, Michael
Defense Systems Division
Biological Sciences and Systems Dept.
General Motors Technical Center
Warren, Michigan

Burnstine, Murray
Research on Fatal Highway Collisions
Dept. of Legal Medicine
Medical School, Harvard University
Cambridge, Mass.

Cain, Dr. Clarence
Lockheed Aircraft Corp.
Missiles & Space Division
Department 61-13
Sunnyvale, Calif.

Campbell, Dr. B. J.
Assistant Director, Automotive
Crash Injury Research of
Cornell University
316 East 61st Street
New York 21, N. Y.

Cappel, Klaus
The Franklin Institute
Philadelphia, Pa.

Carroll, John
Human Factors Division
Bureau of Flight Safety
Civil Aeronautics Board
Washington 25, D. C.

Carter, Dr. R. L.
North American Aviation, Inc.
Columbus Division, Dept. 56
4300 East Fifth Street
Columbus 16, Ohio

Cassidy, William B.
Chance Vought Corp.
3415 Timberview Road
Dallas, Texas

Chandler, Richard F.
Biodynamics Branch, Aeromedical Field Laboratory
Holloman Air Force Base
New Mexico

Clark, Dr. Carl C.
Manager, Life Sciences Department
Martin Company
Baltimore 3, Md.

Coermann, Dr. Rolf
Bioacoustic Laboratory
Aerospace Medical Laboratory
Wright-Patterson Air Force Base
Dayton, Ohio

Creer, Brent
NASA Ames Research Center
Palo Alto, Calif.

Culver, Dr. B. D.
Manager, Life Sciences
Aerojet General Corp.
Box 296
Azusa, Calif.

Derbyshire, George
National Academy of Sciences
2101 Constitution Avenue
Washington 25, D. C.

Dille, J. Robert, M. D.
Civil Aeromedical Research Institute
Federal Aviation Agency
Oklahoma City, Okla.

Doyle, Bernard
Chief, Human Factors Division
Bureau of Flight Safety
Civil Aeronautics Board
Washington 25, D. C.

Drum, Lt. D. E.
Biodynamics Research
Naval Medical Research Institute
Bethesda 14, Md.

Earley, John C.
Civil Aeromedical Research Institute
Federal Aviation Agency
Oklahoma City, Okla.

Foppe, Frank
Dynamicist
Stanley Aviation Corporation
2501 Dallas Street
Denver 8, Colo.

Franks, Dr. Wilbur R.
Group Captain
RCAF Institute of Aviation Medicine
1107 Avenue Road
Toronto 12, Ontario, Canada

Freeman, Howard E.
Unit 87100
Vought Astronautics Division
Chance Vought Corp.
Dallas, Texas

Friede, Dr. Reinhard L.
Department of Psychiatry
University Hospital
University of Michigan
Ann Arbor, Michigan

Gadd, Charles
Supervisor, Dynamics and Stress Analysis
Electro Mechanics Department
Research Laboratories
General Motors Corporation
Twelve Mile and Mounds Road
Warren, Michigan

Gell, Dr. Charles F.
Chief of Life Sciences
Vought Astronautics
Box 5907
Dallas 22, Texas

Gikas, Thomas
National Academy of Sciences
2101 Constitution Avenue
Washington 25, D. C.

Goddard, Dr. James
Civil Air Surgeon
Federal Aviation Agency
Washington 25, D. C.

Gold, Armand
Life Support Systems
Missile & Space Vehicle Dept.
General Electric Co.
3918 Chestnut Street
Philadelphia 4, Pa.

Goldman, CDR David, MSC USN
Naval Medical Research Institute
Bethesda 14, Md.

Gray, R. F.
Biophysics Division
Aviation Medical Acceleration Lab.
U. S. Naval Air Development Center
Johnsville, Pa.

Gros, Percy J. Jr.
Aeromedical Field Laboratory
Holloman Air Force Base
New Mexico

Gross, Arthur G.
University of California
Los Angeles, California

Guignard, Dr. J.
RAF Institute of Aviation Medicine
Farnborough, Hants., England

Gurdjian, Dr. E. S.
Biomechanics Research Center
College of Engineering & Medicine
Wayne State University
Detroit 2, Mich.

Halstead, Maj. James R.
Armed Forces Institute of Pathology
Washington 25, D. C.

Hardy, Dr. James D.
John B. Pierce Foundation
290 Congress Avenue
New Haven 11, Conn.

Hasbrook, Dr. A. Howard
Chief, Crash Safety Section
Civil Aeromedical Research Institute
Federal Aviation Agency
P. O. Box 1082
Oklahoma City, Okla.

Hay, Capt. H. B.
Commanding Officer
RCAF Institute of Aviation Medicine
1107 Avenue Road
Toronto 12, Ontario, Canada

Haynes, Alex L.
Director of Advanced Products Study Engineering
Engineering Staff
Ford Motor Company
20000 Rotunda Drive
Dearborn, Mich.

Hessberg, Col. Rufus, MC USAF
Bioastronautics Office
Aerospace Systems Command
Andrews Air Force Base, Md.

Hiatt, Dr. Edwin P.
Department of Physiology
Ohio State School of Medicine
Columbus, Ohio

Hirsch, Arthur E.
Personnel Protection Section
Code 758
David Taylor Model Basin
Washington 7, D. C.

Holcomb, Dr. Galen A.
Chief, Human Factors, Reliability, & Maintainability Engineering
Stanley Aviation Corp.
2501 Dallas Street
Denver 8, Colo.

Huelke, Dr. Don
Department of Anatomy
College of Medicine
University of Michigan
Ann Arbor, Mich.

Hyde, Dr. A. S.
Acceleration Section
Aerospace Medical Laboratory
Wright-Patterson Air Force Base
Dayton, Ohio

Kirby, Dr. William H.
Room 222, Terminal Ballistic Laboratory
Ballistic Research Laboratories
Aberdeen Proving Ground, Md.

Konecci, Dr. Eugene
Life Sciences Section
Missiles and Space Systems Engineering
Douglas Aircraft Co.
Santa Monica, Calif.

Kornhauser, Murray
Life Support Systems
Missile and Space Vehicle Dept.
The General Electric Co.
3918 Chestnut Street
Philadelphia 4, Pa.

Kraft, Merwyn A.
Research Coordinator
Flight Safety Foundation
468 Park Avenue South
New York 16, N. Y.

Lange, Dr. William
Wayne State University
3919 John R
Detroit 1, Mich.

Lansberg, Dr. M. P.
National Aeromedical Center
Soesterberg, The Netherlands

LeVora, Norman W.
Martin Company
3535 Western Union Avenue
Littleton, Colo.

Libber, Dr. Leonard
Head, Physiology Branch
Office of Naval Research
Building T-3, Room 2704
17th and Constitution Ave., N. W.
Washington 25, D. C.

Lindveit, Dr. Earl W.
General Motors Corporation
1625 I Street, N. W.
Washington, D. C.

Lissner, Dr. Herbert R.
Coordinator, Biomechanics Research Center
College of Engineering & Medicine
Wayne State University
Detroit, Mich.

Lombard, Dr. C. F.
Life Sciences
Norair Division
Northrop Corporation
Hawthorne, Calif.

Mayo, Alfred M.
Manager Advanced Systems Dept.
Astronautics Division
Chance Vought Corp.
Box 6267
Dallas 22, Texas

McCourt, Francis
Assistant Director for Aviation
U. S. Army Transportation Command
Fort Eustis, Va.

Moore, John
American Safety Equipment Corp.
261 Madison Avenue
New York, N. Y.

Moseley, Dr. Alfred
Research on Fatal Highway Collisi
Dept. of Legal Medicine
Harvard University Medical School
Cambridge, Massachusetts

Parker, F. A.
General Electric Company
2044 Sussex Blvd.
Broomall, Pa.

Patrick, Dr. L.
Biomechanics Research Center
Wayne State University
Detroit, Mich.

Payne, Peter R.
Frost Engineering Development Corp.
830 South Lipan Street
Denver 23, Colo.

Perey, Dr. Olof
Lasarettet
Orebro, Sweden

Pesman, G. J.
National Aeronautics & Space Administration
108 Linda Drive
wport News, Va.

Pine, Bruce W.
Biological Sciences and Systems Dept.
Defense Systems Division
General Motors Technical Center
Warren, Mich.

Poisson, William H.
Manager, Planning and Evaluation
Fiber Marketing Department
Allied Chemical Corporation
261 Madison Avenue
New York 16, N. Y.

Rathert, George
NASA Ames Research Center
Palo Alto, Calif.

Rhein, Capt. Leroy W.
Aeromedical Field Laboratory
Holloman Air Force Base
New Mexico

Ross, Malcolm D.
logical Sciences and Systems Dept.
fense Systems Division
General Motors Technical Center
Warren, Mich.

Rothe, Victor E.
Manager, Aviation Crash Injury Research
Flight Safety Foundation
2871 Sky Harbor Blvd.
Phoenix, Arizona

Ryan, Dr. James J.
Automotive Safety
Mechanical Engineering Dept.
University of Minnesota
Minneapolis, Minn.

Scheich, Dr. Wilhelm
Human Factors Group
Space and Information Systems Div.
North American Aviation Inc.
Downey, Calif.

Schneider, Capt. D. J.
Aviation Crash Injury Research
Phoenix, Arizona

Shafer, J. R., Maj.
Headquarters Air Force Systems Command
Office of the Assistant for Bioastronautics
Andrews Air Force Base,
Washington 25, D. C.

Shapland, Dr. D. J.
Supervisor of Research & Development
Stanley Aviation Corp.
2501 Dallas Street
Denver 8, Colo.

Shepard, T. W. Jr.
Project Engineer, Flight Safety
Chance Vought Corp. Foundation
P. O. Box 5907
Dallas 22, Texas

Smedal, Capt. Harold, MC USN
NASA Ames Research Center
Palo Alto, Calif.

Smith, Capt. F. Kirk, MC USN
Director, Aviation Medical Acceleration Laboratory
U. S. Naval Air Development Center
Johnsville, Pa.

Stapp, Col. John P., MC USAF
U. S. Air Force School of Aviation Medicine
Brooks Air Force Base
San Antonio, Texas

Strickland, Brig. Gen. B. A.
Headquarters Air Force Systems Command
Andrews Air Force Base
Washington 25, D. C.

Swearingen, Dr. John
Civil Aeromedical Research Institute
Federal Aviation Agency
P. O. Box 1082
Oklahoma City, Okla.

Taylor, Dr. Ellis R.
Brooks Air Force Base, Texas

Turnbow, Dr. James W.
Director, Experimental Research
Aviation Crash Injury Research
Flight Safety Foundation
2871 Sky Harbor Blvd.
Phoenix, Arizona

Violette, Médecin Lt. Col. Francois
Centre d'Enseignement de Recherches
de Medecine Aeronautique du
Service de Sante (Air)
5 bis, Avenue de la Porte de Sevres
Paris 15me, France

von Gierke, Dr. Henning
Chief, Bioacoustic Laboratory
Aerospace Medical Laboratory
Wright-Patterson Air Force Base
Dayton, Ohio

Wells, Dr. Gordon J.
Space and Information Systems Division
North American Aviation, Inc.
Downey, California

Wilson, R. R.
Aeronautical Systems Division
Wright-Patterson Air Force Base
Dayton, Ohio

Wilburn, Capt. Carl E. MC USN
Aviation Medical Service
Federal Aviation Agency
Washington 25, D. C.

Wolf, Robert
Director, Automotive Crash Injury Research
Cornell University
316 East 61st Street
New York 21, N. Y.

Yost, William R., Capt.
Headquarters Air Force Systems Command
Andrews Air Force Base
Washington 25, D. C.

A CHRONOLOGICAL BIBLIOGRAPHY ON THE BIOLOGICAL EFFECTS OF IMPACT

Carl C. Clark, Denis Faubert, and Bruce Cooper
Life Sciences Department
Martin Company
Baltimore, Maryland

12876

INTRODUCTION

This document has been compiled as part of a preliminary effort to develop a more complete bibliography on the biological effects of acceleration, in connection with a review of impact acceleration by the Working Group Committee of the Space Science Board of the National Academy of Sciences--National Research Council. This bibliography is a collation of previous works (with further additions) that were distributed to participants of the Symposium. Additions and corrections should be sent to Dr. Carl Clark, Manager, Life Sciences Department, the Martin Company.

Appreciation is expressed to the following who submitted additions: Mr. Merwyn Kraft, Dr. F. Violette, Mr. G. A. Holcomb, Mr. R. Flanagan Gray, Dr. James Ryan, Mr. Murray Burnstine, Mr. Armand Gold, Mr. Murray Kornhauser, Col. Karl H. H[illegible]ton, Dr. John Swearingen, Dr. Herbert Lissner, Dr. E. S. Gurdjian, and Mr. J. B. Behr.

A <u>chronological index,</u> sequenced within each year alphabetically by author, is presented as the principal index with the hope that such a tabulation, with corrections, can attain a completeness impossible for an <u>author index.</u> However, an author index, giving years of the author's publications and in some cases a current address, is presented following the chronological index. For co-authors, the name of the first author as well as the year of the publication is specified. An <u>organization index</u> is also included, although this does not include all of the papers of the other indexes. This index, alphabetical by laboratories or groups from which the publications came, gives the first author and year of each publication. The incompleteness of this preliminary bibliography is apparent.

It is emphasized that this is a preliminary bibliography. In too many cases the original publication was not obtained, nor were journals systematically scanned for articles not referenced in previous bibliographies or in articles which are referenced. The following were of particular use in starting this work.

"Human Tolerance to Rapidly Applied Accelerations: A Summary of the Literature," by A. Martin Eiband, NASA Memo 5-19-59E, 1959.

"A Bibliography of Aviation Medicine," by E. C. Hoff and J. F. Fulton, Springfield, Ill., and Baltimore, 1942. Supplement, by P. M. Hoff, E. C. Hoff, and J. F. Fulton, Washington, 1944.

"Aviation Medicine, An Annotated Bibliography," Vol. I, 1956; Vol. II (1953 Literature), 1959, continuing as "Aerospace Medicine and Biology, An Annotated Bibliography," (Vol. III [1954 Literature], 1960, and Vol. IV [1955 Literature], 1961), by Arnold J. Jacobius, <u>et al.</u>, Science and Technology Division, The Library of Congress, Washington, D. C.

"Bioastronautics, An ASTIA Report Bibliography," 1959, with Supplement in 1960, by the Armed Services Technical Information Agency, Arlington 12, Virginia.

However, this preliminary work does provide a skeleton to which improvements may be added without the duplication of effort required to identify commonly referenced works. The authors express apologies to all those slighted in this preliminary work. Corrections and additions are respectfully requested.

The scope of this bibliography has been fairly broadly interpreted: the article must cover some aspect of the biological effects, observed or predicted, of impact. Certain articles describing impact accelerations but without direct biological observations, particularly those involving measurements with dummies, have been included. Webster's Collegiate Dictionary defines impact as "a striking together; a collision communicating force; act of impinging, as a stream on a vane..." To retain the connotation of severity and abruptness of impact for biological effects, we draw an approximate limit that accelerations of concern shall be greater than 10 g and rates of change of acceleration greater than 20 g per second. Articles discussing biological effects of lesser accelerations or those with slower onsets would be classified as "biological effects of maneuver accelerations," and not included here. Thus the majority of the centrifuge work is excluded.

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3. Frankfort Arsenal Laboratory Division, Philadelphia, Pa.

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4. Office of the Surgeon General

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2. Bureau of Medicine and Surgery

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